⑲ Europäisches Patentamt
European Patent Office
Office européen des brevets

⑪ Publication number: **0 261 789 A1**

⑫ EUROPEAN PATENT APPLICATION

㉑ Application number: 87307258.1

(51) Int. Cl.⁴: **A61B 5/00 , G01N 21/31**

㉒ Date of filing: 17.08.87

㉚ Priority: 18.08.86 US 897664

㊸ Date of publication of application:
30.03.88 Bulletin 88/13

(84) Designated Contracting States:
AT BE CH DE ES FR GB GR IT LI LU NL SE

(71) Applicant: **PHYSIO-CONTROL CORPORATION**
11811 Willows Road
Redmond Washington 98052(US)

(72) Inventor: **Cheung, Peter Wing-Poon**
7865 W. Mercer Way
Mercer Island Washington, 98040(US)
Inventor: **Gauglitz, Karl Frederick**
12113 N.E. 151st Street
Kirkland Washington, 98034(US)
Inventor: **Mason, Lee Radford**
1025 N.W. Honeywood Place
Issaquah Washington, 98027(US)
Inventor: **Prosser, Stephen J.**
5706-172nd Place, S.W.
Lynwood Washington, 98037(US)
Inventor: **Smith, Robert Eugene**
703 Fourth Avenue,S.
Edmonds Washington, 98037(US)
Inventor: **Wagner, Darrell Orvin**
28925-104th Place, S.E.
Monroe Washington, 98272(US)
Inventor: **Hunsaker, Scott W.**
1731 N.E. Naomi Place
Seattle Washington, 98115(US)

(74) Representative: **Arthur, Bryan Edward et al**
Withers & Rogers 4 Dyer's Buildings Holborn
London EC1N 2JT(GB)

(54) Feedback-controlled method and apparatus for processing signals.

(57) A feedback control system is disclosed for use in processing signals employed in pulse transmittance oximetry. The signals are produced in response to light transmitted through, for example, a finger at two different wavelengths. Each signal includes a slowly varying baseline component representing the relatively fixed attenuation of light produced by bone, tissue, skin, and hair. The signals also include pulsatile components representing the attenuation produced by the changing blood volume and oxygen saturation within the finger. The signals are processed by the feedback control system before being converted by an analog-to-digital (A/D) converter (72) for subsequent analysis by a microcomputer (16). The feedback control system includes a controllable offset subtractor (66), a programmable gain amplifier (168), controllable drivers (44) for the light sources (40, 42) and the microcomputer (16). The microcomputer (16) receives signals from the offset subtractor (66), gain amplifier (68), drivers (44) and A/D converter (72) to produce signals that control the function of the subtractor (66) and drivers (44) in the following manner. Normally, the drivers (44) are maintained within a predetermined current range. In the event the microcomputer (16) senses an output from the converter (72) that

EP 0 261 789 A1

is not within a predetermined range, the drive signal is adjusted to produce an acceptable signal. The magnitude of the offset removed by the subtractor (66), as controlled by the microcomputer (16), is maintained at a constant level when the converter (72) output is within a first predetermined range and is a predetermined function of the converter (72) output when that output falls within a second predetermined range.

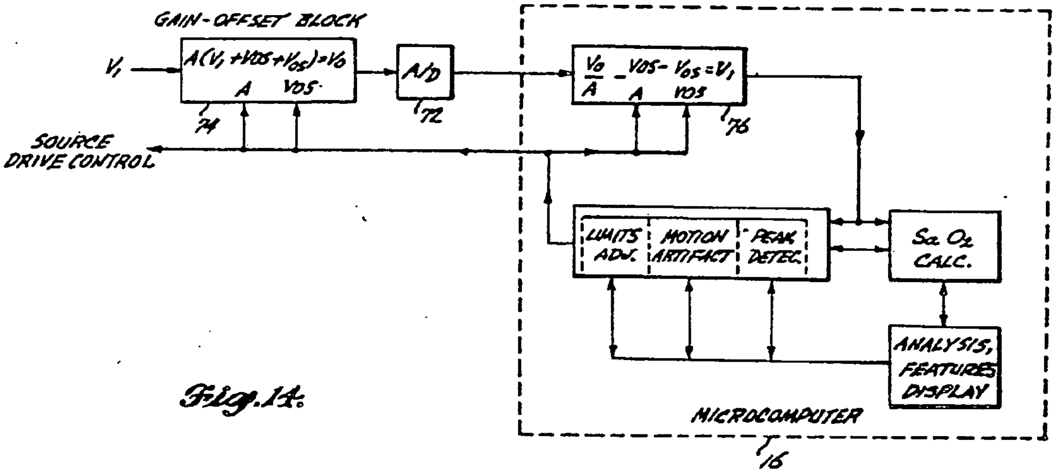

Fig. 14.

FEEDBACK-CONTROLLED METHOD AND APPARATUS FOR PROCESSING SIGNALS USED IN OXIMETRY

Background of the Invention

This invention relates to oximetry and, more particularly, to signal-processing techniques employed in oximetry.

The arterial oxygen saturation and pulse rate of an individual may be of interest for a variety of reasons. For example, in the operating room up-to-date information regarding oxygen saturation can be used to signal changing physiological factors, the malfunction of anaesthesia equipment, or physician error. Similarly, in the intensive care unit, oxygen saturation information can be used to confirm the provision of proper patient ventilation and allow the patient to be withdrawn from a ventilator at an optimal rate.

In many applications, particularly including the operating room and intensive care unit, continual information regarding pulse rate and oxygen saturation is important if the presence of harmful physiological conditions is to be detected before a substantial risk to the patient is presented. A noninvasive technique is also desirable in many applications, for example, when a home health care nurse is performing a routine check-up, because it increases both operator convenience and patient comfort. Pulse transmittance oximetry is addressed to these problems and provides noninvasive, continual information about pulse rate and oxygen saturation. The information produced, however, is only useful when the operator can depend on its accuracy. The method and apparatus of the present invention are, therefore, directed to the improved accuracy of such information without undue cost.

As will be discussed in greater detail below, pulse transmittance oximetry basically involves measurement of the effect arterial blood in tissue has on the intensity of light passing therethrough. More particularly, the volume of blood in the tissue is a function of the arterial pulse, with a greater volume present at systole and a lesser volume present at diastole. Because blood absorbs some of the light passing through the tissue, the intensity of the light emerging from the tissue is inversely proportional to the volume of blood in the tissue. Thus, the emergent light intensity will vary with the arterial pulse and can be used to indicate a patient's pulse rate. In addition, the absorption coefficient of oxyhemoglobin (hemoglobin combined with oxygen, HbO_2) is different from that of deoxygenated hemoglobin (Hb) for most wavelengths of light. For that reason, differences in the amount of light absorbed by the blood at two different wavelengths can be used to indicate the hemoglobin oxygen saturation, % SaO_2 (OS), which equals $([HbO_2]/([Hb] + [HbO_2])) \times 100\%$. Thus, measurement of the amount of light transmitted through, for example, a finger can be used to determine both the patient's pulse rate and hemoglobin oxygen saturation.

As will be appreciated, the intensity of light transmitted through a finger is a function of the absorption coefficient of both "fixed" components, such as bone, tissue, skin, and hair, as well as "variable" components, such as the volume of blood in the tissue. The intensity of light transmitted through the tissue, when expressed as a function of time, is often said to include a baseline component, which varies slowly with time and represents the effect of the fixed components on the light, as well as a periodic pulsatile component, which varies more rapidly with time and represents the effect that changing tissue blood volume has on the light. Because the attenuation produced by the fixed tissue components does not contain information about pulse rate and arterial oxygen saturation, the pulsatile signal is of primary interest. In that regard, many of the prior art transmittance oximetry techniques eliminate the so-called "DC" baseline component from the signal analyzed.

For example, in U.S. Patent No. 2,706,927 (Wood) measurements of light absorption at two wavelengths are taken under a "bloodless" condition and a "normal" condition. In the bloodless condition, as much blood as possible is squeezed from the tissue being analyzed. Then, light at both wavelengths is transmitted through the tissue and absorption measurements made. These measurements indicate the effect that all nonblood tissue components have on the light. When normal blood flow has been restored to the tissue, a second set of measurements is made that indicates the influence of both the blood and nonblood components. The difference in light absorption between the two conditions is then used to determine the average oxygen saturation of the tissue, including the effects of both arterial and venous blood. As will be readily apparent, this process basically eliminates the DC, nonblood component from the signal that the oxygen saturation is extracted from.

For a number of reasons, however, the Wood method fails to provide the necessary accuracy. For example, a true bloodless condition is not practical to obtain. In addition, efforts to obtain a bloodless condition, such as by squeezing the tissue, may result in a difference light transmission path for the two conditions. In addition to problems with accuracy, the Wood approach is both inconvenient and time consuming.

A more refined approach to pulse transmittance oximetry is disclosed in U.S. Patent No. 4,086,915 (Kofsky et al.). The Kofsky et al. reference is of interest for two reasons. First, the technique employed automatically eliminates the effect that fixed components in the tissue have on the light transmitted therethrough, avoiding the need to produce bloodless tissue. More particularly, as developed in the Kofsky et al. reference from the Beer-Lambert law of absorption, the derivatives of the intensity of the light transmitted through the tissue at two different wavelengths, when multiplied by predetermined pseudocoefficients, can be used to determine oxygen saturation. Basic mathematics indicate that such derivatives are substantially independent of the D C component of the intensity. The pseudocoefficients are determined through measurements taken during a calibration procedure in which a patient first respires air having a normal oxygen content and, later, respires air of a reduced oxygen content. As will be appreciated, this calibration process is at best cumbersome.

The second feature of the Kofsky et al. arrangement that is of interest is its removal of the DC component of the signal prior to being amplified for subsequent processing. More particularly, the signal is amplified to allow its slope (i.e., the derivative) to be more accurately determined. To avoid amplifier saturation, a portion of the relatively large DC component of the signal is removed prior to amplification. To accomplish this removal, the signal from the light detector is applied to the two inputs of a differential amplifier as follows. The signal is directly input to the positive terminal of the amplifier. The signal is also passed through a low-resolution A/D converter, followed by a D/A converter, before being input to the negative terminal of the amplifier. The A/D converter has a resolution of approximately 1/10 that of the input signal. For example, if the signal is at 6.3 volts, the output of the A/D converter would be 6 volts. Therefore, the output of the converter represents a substantial portion of the signal, which typically can be used to approximate the DC signal level. Combination of that signal with the directly applied detector signal at the amplifier produces an output that can be used to approximate the AC signal. As will be readily appreciated, however, the process may be relatively inaccurate because the output of the A/D converter is often a poor indicator of the DC signal.

U.S. Patent No. 4,167,331 (Nielson) discloses another pulse transmittance oximeter. The disclosed oximeter is based upon the principle that the absorption of light by a material is directly proportional to the logarithm of the light intensity after having been attenuated by the absorber, as derived from the Beer-Lambert law. The oximeter employs light-emitting diodes (LEDs) to produce light at red and infrared wavelengths for transmission through tissue. A photosensitive device responds to the light produced by the LEDs and attenuated by the tissue, producing an output current. That output current is amplified by a logarithmic amplifier to produce a signal having AC and DC components and containing information about the intensity of light transmitted at both wave-lengths. Sample-and-hold circuits demodulate the red and infrared wavelength signals. The DC components of each signal are then blocked by a series bandpass amplifier and capacitors, eliminating the effect of the fixed absorptive components from the signal. The resultant AC signal components are unaffected by fixed absorption components, such as hair, bone, tissue, skin. An average value of each AC signal is then produced. The ratio of the two averages is then used to determine the oxygen saturation from empirically determined values associated with the ratio. The AC components are also used to determine the pulse rate.

Another reference addressed to pulse transmittance oximetry is U.S. Patent No. 4,407,290 (Wilber). In that reference, light pulses produced by LEDs at two different wavelengths are applied to, for example, an earlobe. A sensor responds to the light transmitted through the earlobe, producing a signal for each wavelength having a DC and AC component resulting from the presence of constant and pulsatile absorptive components in the earlobe. A normalization circuit employs feedback to scale both signals so that the DC nonpulsatile components of each are equal and the offset voltages removed. Decoders separate the two signals, so controlled, into channels A and B where the DC component from each is removed. The remaining AC components of the signals are amplified and combined at a multiplexer prior to analog-to-digital (A/D) conversion. Oxygen saturation is determined by a digital processor in accordance with the following relationship:

$$OS = \frac{X_1 R(\lambda_1) + X_2 R(\lambda_2)}{X_3 R(\lambda_1) + X_4 R(\lambda_2)}$$

wherein empirically derived data for the constants X_1, X_2, X_3 and X_4 is stored in the processor.

European Patent Application No. 83304939.8 (New,Jr. et al.) discloses an additional pulse transmittance oximeter. Two LEDs expose a body member, for example, a finger, to light having red and infrared wavelengths, with each LED having a one-in-four duty cycle. A detector produces a signal in response that is then split into two channels. The one-in-four duty cycle allows negatively amplified noise signals to be integrated with positively amplified signals including the detector response and noise, thereby eliminating the effect of noise on the signal produced. The resultant signals include a substantially constant DC component and a pulsatile AC component. To improve the accuracy of a subsequent analog-to-digital (A/D) conversion, a fixed DC value is subtracted from the signal prior to the conversion. This level is then added back in by a microprocessor after the conversion. Logarithmic analysis is avoided by the micropressor in the following manner. For each wavelength of light transmitted through the finger, a quotient of the pulsatile component over the constant component is determined. The ratio of the two quotients is then determined and fitted to a curve of independently derived oxygen saturations. To compensate for the different transmission characteristics of different patients' fingers, an adjustable drive source for the LEDs is provided. In addition, an apparatus for automatically calibrating the device is disclosed.

Prior art oximeters have, however, not always employed signal-processing techniques that are adequate to provide maximum resolution of the signal received for analysis. As a result, the accuracy of oxygen saturation and pulse rate determinations made by the oximeter may suffer. The disclosed invention addresses this problem and improves the accuracy previously attainable in the art of oximetry.

Summary of the Invention

The present invention discloses an apparatus for processing signals produced by a sensor that contain information about the oxygen saturation of arterial blood flowing in tissue. The apparatus includes an offset subtractor for subtracting a controlled portion of the sensor signal from that signal. The offset subtractor produces an output substantially equal to the portion of the sensor signal remaining after the controlled portion has been subtracted therefrom. The system also includes a controller, coupled to the offset subtractor, which receives the output of the offset subtractor and produces a subtraction control signal dependent upon that output. The subtraction control signal is transferred to the offset subtractor and determines the magnitude of the controlled portion of the signal subtracted thereby. An analyzer receives the output of the offset subtractor and produces an indication of the oxygen saturation of the arterial blood.

In accordance with a particular aspect of the invention, the controlled portion of the detector signal subtracted is held constant when the absolute value of the offset subtractor output is less than a first predetermined level. When the absolute value of the offset subtractor output falls within a predetermined range above that level, however, a subtraction control signal is produced indicating that the offset subtractor is to adjust the magnitude of the controlled portion by an amount proportional to the magnitude of the offset subtractor output.

When the absolute value of the offset subtractor output exceeds a second predetermined level, a subtraction control signal is produced indicating that the offset subtractor is no longer able to adjust the controlled portion of the signal to be subtracted. Preferably, the controlled portion subtracted from the detector signal by the offset subtractor is initialized at a predetermined value.

In accordance with another aspect of the invention, the system further includes a controllable gain amplifier for amplifying the output of the offset subtractor by a controlled gain. The amplifier produces an output that is substantially equal to the product of the offset subtractor output and the gain. The controller produces an amplifier control signal that is received by the amplifier, which adjusts the controlled gain in response thereto.

In accordance with a further aspect of the invention, the controller produces a sensor control signal to which said sensor responds. The controller establishes the sensor control signal at a level sufficient to cause the sensor signal to fall within a predetermined sensor signal range.

In accordance with further aspects of this invention, a differential current-to-voltage amplifier amplifies the sensor signal before it is received by the offset subtractor. An analog-to-digital converter also converts the output of the controllable-gain amplifier into a digital format for analysis. The analyzer removes the gain and adds the controlled portion back to the amplifier output before producing the indication of oxygen saturation.

As will be appreciated, the disclosed invention also includes an oximeter employing the apparatus described above in conjunction with a sensor. The sensor includes a light source that responds to a control signal from the controller and illuminates the tissue. The intensity of the illumination is determined by the control signal. A detector included in the sensor responds to the illumination of the tissue by producing a signal that contains information about the oxygen saturation of the arterial blood. A red optical filter may be included to filter the light received by the detector.

As will be appreciated, the disclosed invention includes the method of processing signals employed by the apparatus discussed above to determine the oxygen saturation of arterial blood flowing in tissue. In a basic form, the method includes the steps of subtracting from the sensor signal a controlled portion of the signal in response to a subtraction control signal. A subtraction output is produced that substantially equals the portion of the sensor signal remaining after the controlled portion has been subtracted therefrom. A subtraction control signal is also produce, dependent on the subtraction output in a manner indicating the desired adjustment in the controlled portion subtracted from the sensor signal.

Brief Description of the Drawings

The invention can best be understood by reference to the following portion of the specification taken in conjunction with the accompanying drawings in which:

FIGURE 1 is a block diagram of an oximeter including a sensor, input/output (I/O) circuit, microcomputer, alarm, displays, power supply, and keyboard; FIGURE 2 is a block diagram illustrating the transmission of light through an absorptive medium; FIGURE 3 is a block diagram illustrating the transmission of light through the absorptive medium of FIGURE 2 wherein the medium is broken up into elemental components; FIGURE 4 is a graphical comparison of the incident light intensity to the emergent light intensity as modeled in FIGURE 2; FIGURE 5 is a graphical comparison of the specific absorption coefficients for oxygenated hemoglobin and deoxygenated hemoglobin as a function of the wavelength of light transmitted therethrough; FIGURE 6 is a block diagram illustrating the transmission of light through a block model of the components of a finger; FIGURE 7 is a graphical comparison of empirically derived oxygen saturation measurements with a variable that is measurable by the oximeter; FIGURE 8 is a schematic illustration of the transmission of light at two wavelengths through a finger in accordance with the invention; FIGURE 9 is a graphical plot as a function of time of the transmittance of light at the red wavelength through the finger; FIGURE 10 is a graphical plot as a function of time of the transmission of infrared light through the finger; FIGURE 11 is a more detailed schematic of the I/O circuit illustrated in the system of FIGURE 1; FIGURE 12 is a schematic diagram of a conventional current-to-voltage amplifier circuit; FIGURE 13 is a schematic diagram of a differential current-to-voltage preamplifier circuit included in the I/O circuit of FIGURE 1; FIGURE 14 is a functional block diagram illustrating the basic operation of the feedback control system constructed in accordance with this invention; FIGURE 15 is a graphical representation of the possible ranges of I/O circuit output, showing the desired response of the I/O circuit and microcomputer at each of the various possible ranges; FIGURE 16 is a block diagram of a portion of an interrupt level software routine included in the microcomputer illustrated in FIGURE 1; FIGURES 17 through 19 are more detailed block diagrams of the interrupt level routine depicted in FIGURE 16; FIGURE 20 is a graphical representation of the possible ranges of current supplied to the sensor, showing the desired response of the I/O circuit and microcomputer at each of the various possible ranges as a function of sensor output; FIGURES 21 through 23 are further detailed block diagrams of the interrupt level routine depicted in FIGURE 16; FIGURE 24 is a block diagram of reconstruction software included in the microcomputer illustrated in FIGURE 1; FIGURE 25 illustrates a calibrated offset table stored in the microcomputer for use in adjusting the operation of the I/O circuit; and FIGURE 26 is a more complete schematic diagram of the microcomputer illustrated in FIGURE 1.

Detailed Description

Referring to the overall system block diagram shown in FIGURE 1, a pulse transmittance oximeter 10 employing this invention includes a sensor 12, input/output (I/O) circuit 14, microcomputer 16, power source 18, display 20, keyboard 22 and alarm 24. Before discussing these elements in detail, however, an outline of the theoretical basis of pulse transmittance oximetry as practiced by the oximeter of FIGURE 1 is provided.

An understanding of the relevant theory begins with a discussion of the Beer-Lambert law. This law governs the absorption of optical radiation by homogenous absorbing media and can best be understood with reference to FIGURES 2 and 3 in the following manner.

As shown in FIGURE 2, incident light having an intensity I_O impinges upon an absorptive medium 26. Medium 26 has a characteristic absorbance factor A that indicates the attenuating affect medium 26 has on the incident light. Similarly, a transmission factor T for the medium is defined as the reciprocal of the absorbance factor, I/A. The intensity of the light I_1 emerging from medium 26 is less than I_O and can be expressed functionally as the product TI_O. With medium 26 divided into a number of identical components, each of unit thickness (in the direction of light transmission) and the same transmission factor T, the effect of medium 26 on the incident light I_O is as shown in FIGURE 3.

There, medium 26 is illustrated as consisting of three components 28, 30, and 32. As will be appreciated, the intensity I_1 of the light emerging from component 28 is equal to the incident light intensity I_O multiplied by the transmission factor T. Component 30 has a similar effect on light passing therethrough. Thus, because the light incident upon component 30 is equal to the product TI_O, the emergent light intensity I_2 is equal to the product TI_1 or T^2I_O. Component 32 has the same effect on light and, as shown in FIGURE 3, the intensity of the emergent light I_3 for the entire medium 26 so modeled is equal to the product TI_2 or T^3I_O. If the thickness d of medium 26 is n unit lengths, it can be modeled as including n identical components of unit thickness. It will then be appreciated that the intensity of light emerging from medium 26 can be designated I_n and the product is equal to T^nI_O. Expressed as a function of the absorbance constant A, I_n can also be written as the product $(1/A^n)I_O$.

From the preceding discussion, it will be readily appreciated that the absorptive effect of medium 26 on the intensity of the incident light I_O is one of exponential decay. Because A may be an inconvenient base to work with, $I_n$can be rewritten as a function of a more convenient base, b, by recognizing that A^n is equal to $b^{\alpha n}$, where $^{\alpha}$ is the absorbance of medium 26 per unit length. The term α is frequently referred to as the relative extinction coefficient and is equal to $\log_b A$.

Given the preceding discussion, it will be appreciated that the intensity of the light I_n emerging from medium 26 can be expressed in base 10 (where $\alpha = \alpha_1$)as $I_O10^{-\alpha_1 n}$, or in base e (where $=^{\alpha_2}$)as $I_Oe^{-\alpha_2 n}$. The effect that the thickness of medium 26 has on the emergent light intensity I_n is graphically depicted in FIGURE 4. If the light incident upon medium 26 is established as having unit intensity, FIGURE 4 also represents the transmission factor T of the entire medium as a function of thickness.

The discussion above can be applied generally to the medium 26 shown in FIGURE 2 to produce:

$$I_1 = I_Oe^{-\alpha d} \qquad (1)$$

where I_1 is the emergent light intensity, I_O is the incident light intensity, α is the absorbance coefficient of the medium per unit length, d is the thickness of the medium in unit lengths, and the exponential nature of the relationship has arbitrarily been expressed in terms of base e. Equation (1) is commonly referred to as the Beer-Lambert law of exponential light decay through a homogenous absorbing medium.

With this basic understanding of the Beer-Lambert law, a discussion of its application to the problems of pulse rate and hemoglobin oxygen saturation measurement is now presented. As shown in FIGURE 5, the absorption coefficients for oxygenated and deoxygenated hemoglobin are different at every wavelength, except an isobestic wavelength. Thus, it will be appreciated that if a person's finger is exposed to incident light and the emergent light intensity measured, the difference in intensity between the two, which is the amount of light absorbed, contains information relating to the oxygenated hemoglobin content of the blood in the finger. The manner in which this information is extracted from the Beer-Lambert law is discussed below. In addition, it will be appreciated that the volume of blood contained within an individual's finger varies with the individual's pulse. Thus, the thickness of the finger also varies slightly with each pulse, creating a changing path length for light transmitted through the finger. Because a longer lightpath allows additional light to be absorbed, time-dependent information relating to the difference between the incident and emergent light intensities can be used to determine the individual's pulse. The manner in which this information is extracted from the Beer-Lambert law is also discussed below.

As noted in the preceding paragraph, information about the incident and emergent intensities of light transmitted through a finger can be used to determine oxygen saturation and pulse rate. The theoretical basis for extracting the required information, however, is complicated by several problems. For example, the precise intensity of the incident light applied to the finger is not easily determined. Thus, it may be necessary to extract the required information independently of the intensity of the incident light. Further, because the changing volume of blood in the finger, and, hence, thickness of the lightpath therethrough, are not exclusively dependent upon the individual's pulse, it is desirable to eliminate the changing path length as a variable from the computations.

The manner in which the Beer-Lambert law is refined to eliminate the incident intensity and path length as variables is as follows. With reference to FIGURE 6, a human figure is modeled by two components 34 and 36, in a manner similar to that shown in FIGURE 3. Baseline component 34 models the unchanging absorptive elements of the finger. This component includes, for example, bone, tissue, skin, hair, and baseline venous and arterial blood and has a thickness designated d and an absorbance α.

Pulsatile component 36 represents the changing absorptive portion of the finger, the arterial blood volume. As shown, the thickness of this component is designated Δd, representing the variable nature of the thickness, and the absorbance of this arterial blood component is designated $\alpha_A$representing the arterial blood absorbance.

As will be appreciated from the earlier analysis with respect to FIGURE 3, the light I_1 emerging from component 34 can be written as a function of the incident light intensity I_0 as follows:

$I_1 = I_0 e^{-\alpha d}$ (2)

Likewise, the intensity of light I_2 emerging from component 36 is a function of its incident light intensity I_1,and:

$I_2 = I_1 e^{-\alpha_A \Delta d}$ (3)

Substitution of the expression for I_1 developed in equation (2) for that used in equation (3), when simplified, results in the following expression for the intensity I_2 of light emerging from the finger as a function of the intensity of light I_0incident upon the finger;

$I_2 = I_0 e^{-[\alpha d + \alpha_A \Delta d]}$ (4)

Because our interest lies in the effect on the light produced by the arterial blood volume, the relationship between I_2 and I_1 is of particular interest. Defining the change in transmission produced by the arterial component 36 as $T_{\Delta A}$, we have:

$$T_{\Delta A} = \frac{I_2}{I_1} \qquad (5)$$

Substituting the expressions for I_1 I_2 obtained in equations (2) and (3), respectively, equation (5) becomes:

$$T_{\Delta A} = \frac{I_0 e^{-[\alpha d + \alpha_A \Delta d]}}{I_0 e^{-\alpha d}} \qquad (6)$$

It will be appreciated that I_0 term can be cancelled from both the numerator and denominator of equation (6)d, thereby eliminating the input light intensity as a variable in the equation. With equation (6) fully simplified, the change in arterial transmission can be expressed as:

$T_{\Delta A} = e^{-\alpha_A \Delta d}$ (7)

A device employing this principle of operation is effectively self-calibrating, being independent of the incident light intensity I_0.

At this point, a consideration of equation (7) reveals that the changing thickness of the finger, Δd, produced by the changing arterial blood volume still remains as a variable. The Δd variable is eliminated in the following manner. For convenience of expression, the logarithms of the terms in equation (7) are produced with respect to the same base originally employed in equation (1). Thus, equation (7) becomes:

$\ln T_{\Delta A} = \ln(e^{-\alpha_A \Delta d}) = -\alpha_A \Delta d$ (8)

A preferred technique for eliminating the Δd variable utilizes information drawn from the change in arterial transmission experienced at two wavelengths.

The particular wavelengths selected are determined in part by consideration of a more complete expression of the arterial absorbance α_A :

$\alpha_A = (\alpha_{OA})(OS)-(\alpha_{DA})(1-OS)$ (9)

where α_{OA} is the oxygenated arterial absorbance, α_{DA} is the deoxygenated arterial absorbance, and OS is the hemoglobin oxygen saturation of the arterial blood volume. As will be appreciated from FIGURE 5, α_{OA} and α_{DA} are substantially unequal at al light wavelengths in the red and near-infrared wavelength regions except for an isobestic wavelength occurring at approximately 805 nanometers. With an arterial oxygen saturation OS of approximately 90 percent, it will be appreciated from equation (9) that the arterial absorbance α_A is 90 percent attributable to the oxygenated arterial absorbance α_{OA} and 10 percent attributable to the deoxygenated arterial absorbance α_{DA}. At the isobestic wavelength, the relative contribution of these two coefficients to the arterial absorbance α_A is of minimal significance in that both α_{OA} and α_{DA} are equal. Thus, a wavelength roughly approximating the isobestic wavelength of the curves illustrated in FIGURE 5 is a convenient one for use in eliminating the change in finger thickness Δd attributable to arterial blood flow.

A second wavelength is selected at a distance from the appoximately isobestic wavelength that is sufficient to allow the two signals to be easily distinguished. In addition, the relative difference of the oxygenated and deooxygenated arterial absorbances at this wavelength is more pronounced. In light of the foregoing considerations, it is generally preferred that the two wavelengths selected fall within the red and infrared regions of the electromagnetic spectrum.

The foregoing information, when combined with equation (8) is used to produce the following ratio:

$$\frac{\ln T_{\Delta AR}}{\ln T_{\Delta AIR}} = \frac{-\alpha_A \Delta d @ \lambda_R}{-\alpha_A \Delta d @ \lambda_{IR}} \qquad (10)$$

where $T_{\Delta AR}$ equals the change in arterial transmission of light at the red wavelength λ_R and $T_{\Delta AIR}$ is the change in arterial transmission at the infrared wavelength λ_{IR}. It will be appreciated that if two sources are positioned at substantially the same location on the finger, the length of the lightpath through the finger is substantially the same for the light emitted by each. Thus, the change in the lightpath resulting from arterial blood flow, Δd, is substantially the same for both the red and infrared wavelength sources. For that reason, the Δd term in the numerator and denominator of the right-hand side of equation (10) cancel, producing:

$$\frac{\ln T_{\Delta AR}}{\ln T_{\Delta AIR}} = \frac{\alpha_A @ \lambda_R}{\alpha_A @ \lambda_{IR}} \qquad (11)$$

As will be appreciated, equation (11) is independent of both the incident light intensity I_0 and the change in finger thickness Δ d attributable to arterial blood flow. The foregoing derivations form the theoretical basis of pulse oximetry measurement. Because of the complexity of the physiological process, however, the ratio indicated in equation (11) does not directly provide an accurate measurement of oxygen saturation. The correlation between the ratio of equation (11) and actual arterial blood gas measurement is, therefore, relied on to produce an indication of the oxygen saturation. Thus, if the ratio of the arterial absorbance at the red and infrared wavelengths can be determined, the oxygen saturation of the arterial blood flow can be extracted from independently derived, empirical calibration curves in a manner independent of I_0 and Δ d.

For simplicity, a measured ratio R_{OS} is defined from equation (11) as:

$$\text{Ratio} = R_{OS} = \frac{\alpha_A @ \lambda_R}{\alpha_A @ \lambda_{IR}} \qquad (12)$$

It is this measured value for R_{OS} that is plotted on the x-axis of independently derived oxygen saturation curves, as shown in FIGURE 7 and discussed in greater detail below, with the hemoglobin oxygen saturation being referenced on the y-axis.

Measurement of the arterial absorbances at both wavelengths is performed in the following manner. As shown in FIGURE 8, a detector 38 placed on the side of a finger opposite red and infrared wavelength light sources 40 and 42 receives light at both wavelengths transmitted through the finger. As shown in FIGURE 9, the received red wavelength light intensity, plotted as a function of time, varies with each pulse, and has high and low values R_H and R_L, respectively. R_L occurs substantially at systole, when arterial blood volume is at its greatest; while R_H occurs substantially at diastole, when the arterial blood volume is lowest. From the earlier discussion of the exponential light decay through homogeneous media, it will be appreciated that:

$R_L = I_0 e^{-[\alpha d + \alpha_A \Delta d]}{}_{\alpha \lambda_R}$ (13)

Similarly: $R_H = I_0 e^{-\alpha d}{}_{\alpha \lambda_R}$ (14)

Taking the ratio of equations (13) and (14) and simplifying, we have:

$$\frac{R_L}{R_H} = \frac{I_0 e^{-[\alpha d + \alpha_A \Delta d]}}{I_0 e^{-\Delta d}} @ \lambda_R = e^{-\alpha_A \Delta d} @ \lambda_R \qquad (15)$$

Taking the logarithm of both sides of equation (15) produces:
$\ln (R_L/R_H) = \ln (e^{-\alpha_A \Delta d}) @ \lambda_R = - \alpha_A \Delta d \; \alpha \; \lambda_R$ (16)
As will be readily appreciated, similar expression can be produced for the signals representative of the infrared wavelength light received by detector 38. Thus, the mininum light intensity passing through the finger at the infrared wavelength can be written:
$IR_L = I_0 e^{-[\alpha d + \alpha_A \Delta d]} @ \lambda_{IR}$ (17)
Similarly, the maximum light intensity emerging from the finger at the infrared wavelength can be expressed as:
$IR_H = I_0 e^{-\alpha d} @ \lambda_{IR}$ (18)
The ratio of the terms in equations (17) an (18) can be expressed as:

$$\frac{IR_L}{IR_H} = \frac{I_0 e^{-[\alpha d + \alpha_A \Delta d]}}{I_0 e^{-\alpha d}} @ \lambda_{IR} = e^{-\alpha_A \Delta d} @ \lambda_{IR} \qquad (19)$$

Use of logarithms simplifies equation (19) to:
$\ln (IR_L/IR_H) = \ln (e^{-\alpha_A \Delta d}) @ \lambda_{IR} = -\alpha_A \Delta d @ \lambda_{IR}$ (20)
The ratiometric combination of equations (16) and (20) yields:

$$\frac{\ln (R_L/R_H)}{\ln (IR_L/IR_H)} = \frac{-\alpha_A \Delta d @ \lambda_R}{-\alpha_A \Delta d @ \lambda_{IR}} \qquad (21)$$

Because the Δd term in the numerator and denominator of the right-hand side of equation (21) cancel, as do the negative signs before each term, it will be appreciated that equation (21) when combined with equation (12) yields:

$$\text{Ratio} = R_{OS} = \frac{\alpha_A @ \lambda_R}{\alpha_A @ \lambda_{IR}} = \frac{\ln(R_L/R_H)}{\ln(IR_L/IR_H)} = \frac{\ln(R_H/R_L)}{\ln(IR_H/IR_L)} \qquad (22)$$

Thus, by measuring the minimum and maximum emergent light intensities at both the red and infrared wavelengths (R_L, R_H, IR_L, IR_H), a value for the term R_{OS} can be computed. From this, empirically derived calibration curves similar to that shown in FIGURE 7 can be used to determine the oxygen saturation as described in greater detail in conjunction with the discussion of the various components of oximeter 10 that follows. As will be appreciated, the determination of oxygen saturation in this manner differs from prior art techniques, such as that disclosed by Wilber, by performing measurements based upon both the baseline and pulsatile components of the signals.

The first component of oximeter 10 to be discussed is sensor 12. The function of sensor 12 is substantially to provide the desired orientation of light sources 40 and 42, for example, light-emitting diodes (LEDs), and light detector 38 with respect to a suitable portion of a patient's body. For example, the sensor must align LEDs 40 and 42 with detector 38 in a manner such that the path of light from each LED to the detector 38 is substantially the same distance. In addition, the path must traverse a portion of the patient's body through which a usable amount of light is passed, for example, a finger, toe, earlobe, or the nasal septum. Because changes in the lightpath can significantly affect the readings taken, as noted above, the sensor must maintain the position of LEDs 40 and 42 and detector 38 with respect to the transmission path through the patient's skin at all times. Otherwise, signal fluctuations known as motionartifact may be produced. In addition, the sensor should apply only insubstantial pressure to the patient's skin and underlying tissue. Otherwise, normal arterial blood flow upon which the pulse oximeter relies for accurate operation, may be disrupted. Finally, the sensor should be quickly attachable to the patient and should cause no discomfort.

LEDs 40 and 42 are supplied with current by transistor drivers 44 located in the I/O circuit 14, as shown in FIGURE 11. Drivers 44 are controlled by microcomputer 16 to produce current pulses at a 960Hz repetition rate. The duration of each pulse is 70 microseconds and a pulse is supplied to the red wavelength LED 40 first and then to the infrared wavelength LED 42. The voltage drop across scaling resistors in th drivers 44 allows the magnitude of the current pulses to be determined and, thus, maintained in a manner described in greater detail below. LEDs 40 and 42 respond to the current pulses by producing corresponding light pulses transmitted through the finger to detector 38. Detector 38, in turn, produces a signal that includes information about the pulsatile response of the finger to the red and infrared wavelength light, intermixed at the 960Hz LED pulse repetition rate.

In a preferred embodiment of the invention, a red optical filter 45 interrupts the lightpath between the LEDs 40 and 42 and the detector 38, as shown in FIGURE 8. Preferably, filter 45 is a Kodak No. 29 wratten gel filter. Its function is to eliminate the influence of fluorescent light flicker on the oxygen saturation determinate made. As will be appreciated, although the body of sensor 12 may b made of an opaque material that blocks a significant portion of the ambient light, some ambient light may still reach detector 38. Light from the sun and incandescent lamps is substantially continuous. Fluorescent lighting, on the other hand, includes alternating energized and deenergized intervals that form a visually imperceptible flicker. The frequency of the fluorescent light flicker is such that it might influence the signal produced by detector 38 in response to light received from LED 40 at the red wavelength. Thus, the red optical filter 45 is placed over the detector 38 and filters out any fluorescent light present, eliminating the effect its flicker might have on the oxygen saturation determination made.

At the I/O circuit 14, the signal from detector 38 is received by a preamplifier 46. In a preferred embodiment, preamplifier 46 includes a differential current-to-voltage amplifier 48 and a single-ended output amplifier 50. To understand the advantages of using the differential amplifier 48, it may first be helpful to consider the operation of a conventional current-to-voltage amplifier as shown in FIGURE 12. As shown, a current-to-voltage amplifier 52 is substantially comprised of an operational amplifier 54 and gain determination resistor R_F. With a detector 38 connected to the inputs of the amplifier as shown, a current I_D is input to the amplifier upon the detection of suitable wavelength light. The output of amplifier 52 is designated V_O and, as will be appreciated, is equal to the product of the detector current I_D and the gain determination resistor R_F. The primary problem with such a construction is that it also amplifies the external interference noise produced, making the signal extracted less accurate.

Adoption of the differential current-to-voltage amplifier 48, when combined with the single-ended output amplifier 50 as shown in FIGURE 13, however, eliminates this problem. As shown, the differential amplifier 48 produces positive and negative versions of the output, the absolute value of each version being equal to the product of the gain determination resistance R_F and the detector current I_D. These outputs are then supplied to the single-ended output amp 50, which provides unity gain, thus producing an output signal having a magnitude that is twice that of the inputs. An advantage of this arrangement is that external interference noise is cancelled at the single-ended output amplifier 50 by the opposing signs of the two transimpedance amplifier outputs. In addition, twice the signal is produced with the current noise only increasing by a magnitude of 1.414. Therefore, an improved signal-to-noise ratio results.

At this point, the mixed signal indicative of the red and infrared wavelength responses of detector 38 has been amplified and is input to a demodulator 56 to extract the red pulsatile and infrared pulsatile waveforms shown in FIGURES 9 and 10. In a preferred arrangement, the demodulator 56 includes a sample-and-hold (S/H) circuit 60 that responds to the detector signal produced in response to red wavelength light and a sample-and-hold (S/H) circuit 58 that responds to the infrared wavelength response of detector 38. The timing of circuits 58 and 60 is controlled so that each circuit samples the signal input to demodulator 56 during the portion of the signal corresponding to the wavelength to which it responds. In this manner, two signals are reconstructed from the single input to demodulator 56. As noted above, these signals correspond to the red pulsatile signal and infrared pulsatile signals shown in FIGURES 9 and 10.

To remove high-frequency noise from the outputs of circuits 58 and 60, they are input to lowpass filters 62 and 64. In a preferred embodiment, the "red" lowpass filter 62 and "infrared" lowpass filter 64 each include two stages. The first stage of each filter utilizes a fifth-order, monolithic integrated circuit switched capacitor filter because of its low cost, relatively small physical size and accuracy. Since both the "red" and "infrared" signals pass through nearly identical first-stage filters due to monolithic IC matching, their gain and phase frequency responses are matched. The second stage of each filter is a second-order Bessel filter having a slightly higher roll-off frequency than the first stage. This ensures that the first-stage filter is the dominant filter of the two-stage combination, producing the desired filtering accuracy. The second stage then filters the switching noise from the first-stage output.

The filtered red and infrared pulsatile signals are next prepared for conversion and transmission to the microcomputer 16. As will be discussed in greater detail below, this process involves the use of a programmable DC subtractor or offset 66 followed by a programmable gain amplifier 68 having a gain range from approximately one to 256. The appropriately processed signals are combined at multiplexer 70, sampled and held at 71, and converted to digital form by A/D converter 72 for transmission to microcomputer 16.

Before a more complete discussion of the operation of programmable subtractor 66, programmable gain amplifier 68, multiplexer 70, S/H 71, and A/D converter 72 is provided, several details regarding the signals to be transferred to microcomputer 16 should be noted. For example, as shown in FIGURES 9 and 10, the signal produced by detector 30 in response to light at each wavelength includes components that, for convenience, are termed baseline and pulsatile. The baseline component approximates the intensity of light received at detector 38 when only the "fixed" nonpulsatile absorptive component is present in the finger. This component of the signal is relatively constant over short intervals but does vary with nonpulsatile physiological changes or system changes, such as movement of sensor 12 on the finger. Over a relatively long interval this baseline component may vary significantly. As will be appreciated, the magnitude of the baseline component at a given point in time is substantially equal to the level identified in FIGURE 9 as R_H. For convenience, however, the baseline component may be thought of as the level indicated by R_L, with the pulsatile component varying between the values for R_H and R_L over a given pulse. That pulsatile component is attributable to light transmission changes through the finger resulting from blood volume changes in the finger during a cardiac pulse. Typically, the pulsatile component may be relatively small in comparison to the baseline component and is shown out of proportion in FIGURES 9 and 10.

The determination of R_{OS} in accordance with equation (22) requires accurately measured values for both the baseline and pulsatile signal components. Because the pulsatile components are smaller, however, greater care must be exercised with respect to the measurement of these components. As will be readily appreciated, if the entire signal shown in FIGURES 9 and 10, including the baseline and pulsatile components, was amplified and converted to a digital format for use by microcomputer 16, a great deal of the accuracy of the conversion would be wasted because a substantial portion of the resolution would be used to measure the baseline component. For example, with an A/D converter employed having an input range of between +10 and -10 volts, a signal having a baseline component referenced to -10 volts that is four times that of the pulsatile component can be amplified until the baseline component is represented by a 16-volt difference and the pulsatile signal represented by a 4-volt difference. With a 12-bit A/D converter 72, the total signal can be resolved into 4096 components. Therefore, the number of incremental levels representing the pulsatile signal would be approximately 819. If, on the other hand, the baseline component is removed prior to the conversion, the gained pulsatile signal could be resolved into 4096 intervals, substantially improving accuracy.

The disclosed invention employs this technique, as the first half of a construction-reconstruction process, in the manner schematically outlined in FIGURE 14. As shown, an input signal V_1 (corresponding to the signals shown in FIGURES 9 and 10) is received from each filter 62 and 64. V_1 includes both the baseline and pulsatile components discussed above. As will be described, subsequent operations upon V_1 subtract off a substantial "offset voltage" portion of the baseline component, then gain up the remaining substantially pulsatile signal for conversion by A/D converter 72. A digital reconstruction of the original signal is then produced by reversing the process, wherein digitally provided information allows the gain to be removed and the offset voltage added back. This step is necessary because the entire signal, including both the baseline and pulsatile components, is used in the oxygen saturation measurement process.

For purposes of the following discussion, an offset voltage V_{OS}(computed in a manner discussed in greater detail in conjunction with the description of microcomputer software provided below) is defined to be the negative value of the portion of the baseline component to be subtracted. Because some offset error voltage is introduced by the various components of the system, the portion of the signal ultimately attributable to these components V_{os} is also preferably accounted for because it represents an error. As shown in FIGURE 14, a signal construction block 74, corresponding to programmable subtractor 66 and programmable amplifier 68, initially processes the input signal V_1 by adding to it the negatively defined offset voltage V_{OS}. The output of construction block 74, V_O, is defined as follows:

$$V_O = (V_1 + V_{OS} + V_{os})\,A \qquad (23)$$

As will be readily appreciated, V_O is substantially proportional to the pulsatile component V_1, which contains the pulsatile information desired. Without gain A, this pulsatile signal may be relatively small in comparison to the maximum input range of A/D converter 72. To provide good resolution, therefore, the signal is amplified by gain A, which is sufficient to produce a signal occupying a predetermined portion of the A/D converter 72 input range. In this manner, the resolution of the digital conversion is improved by providing a large pulsatile signal for measurement.

If R_H, R_L, IR_H and IR_L, as discussed previously, are to be measured and the oxygen saturation determined, however, the foregoing process must be reversed. As will be appreciated, dividing both sides of the equation (23) by the gain A produces:

$$V_1 + V_{OS} + V_{os} = \frac{V_0}{A} \qquad (24)$$

Restructuring of equation (14) results in:

$$V_1 = \frac{V_0}{A} - V_{OS} - V_{os} \qquad (25)$$

Thus, the original input signal V_1 containing the baseline and pulsatile components can be reconstructed at block 76 by dividing the output of the A/D converter 72, V_O, by the gain A and then subtracting the offset voltages V_{OS} and V_{os}. The reconstruction is preferably performed at microcomputer 16 before oxygen saturation computations are initiated, allowing measurements based on the full signal to be performed. As will be appreciated, to accomplish this, values for V_{OS},V_{os} and A must be supplied to microcomputer 16.

Feedback from microcomputer 16 is also required to maintain the values for V_{OS},V_{os} and gain A at levels appropriate to produce optimal A/D converter 72 resolution. Likewise, as shown in the FIGURE 14, feedback to the source drivers 44 may be used to help optimize the conversion. Proper control requires that the microcomputer continually analyze, and respond to, the offset voltages V_{OS} and V_{os}, gain A, driver currents I_O and the output of A/D converter in a manner to be described next.

Briefly, with reference to FIGURE 15, thresholds L1 and L2 slightly below and above the maximum positive and negative excursions L3 and L4 allowable for the A/D converter 72 input are established and monitored by microcomputer 16 at the A/D converter output. When the magnitude of the signal input to, and output from, A/D converter 72 exceeds either of the thresholds L1 or L2, the drive currents I_D are readjusted to increase or decrease the intensity of light impinging upon the detector 38. In this manner, the A/D converter 72 is not overdriven and the margin between L1 and L3 and between L2 and L4 helps assure this even for rapidly varying signals. An operable voltage margin for A/D converter 72 exists outside of the thresholds, allowing A/D converter 72 to continue operating while the appropriate feedback adjustments to A and V_{OS} are made.

When the signal from A/D converter 72 exceeds positive and negative thresholds L5 or L6, microcomputer 16 responds by signaling the programmable subtractor 66 to increase or decrease the voltage V_{os} being subtracted. This is done through the formation and transmission of an offset code whose magnitude is dependent upon the level of the signal received from converter 72.

The manner in which the various thresholds are established and the relationship of the offset codes to the signal received can be altered to produce substantially any form of control desired. In addition, gain control codes could be established by microcomputer 16 in response to the output of A/D converter 72 to vary the gain of amplifier 68 as a function of converter output. Thus, the arrangement shown in FIGURE 15 is illustrative only and represents the currently preferred embodiment. This embodiment of the construction-reconstruction process will now be discussed in greater detail in conjunction with a portion of the oximeter software stored in the erasable, progammable read-only memory (EPROM) 78 of microcomputer 16. The software defines a program of instructions to be executed by the central processing unit (CPU) 80 of microcomputer 16 and governs the manner in which microcomputer 16 provides servosensor control as well as produces measurement outputs for display.

The first segment of the software to be considered is an interrupt level routine 82, shown in part in FIGURE 16. Interrupts are events generated by a programmable timer, which is included in CPU 80 and is initialized at the power-up of microcomputer 16. An interrupt event "interrupts" the part of he program currently being executed by CPU 80 and transfers control to a new instruction sequence associated with the particular interrupt. When processing associated with the interrupt is completed, the program may be resumed at the point of interruption, or elsewhere, depending on any status changes that may have resulted from processing of the interrupt.

Normal interrupt processing in accordance with routine 82 begins once CPU 80 has completed a number of preliminary routines, including power-up reset, calibration, and miscellaneous test code routines. Microcomputer 16 and the software stored in EPROM 78 are organized to provide real-time processing at the interrupt level routine 82, as well as at the other, prioritized task levels noted below. Processing at the various task levels is prioritized such that the highest priority task ready and waiting to run is given control, in the absence of an interrupt. Thus, a task may be interrupted to execute a higher priority task and then resumed when processing at the higher priority level is completed. Interrupt processing occurs at a priority level above all other tasks.

While the interrupt routine 82 employed may have a number of subroutines controlling various portions of oximeter 10 operation, such as the filtering performed by lowpass filters 62 and 64, only the details of the interrupt period subroutine directly pertinent to servosensor control are shown in FIGURE 16.

As shown at block 84, processing of the interrupt level routine 82 does not begin until calibration is complete. After calibration, a nominal interrupt period zero subroutine 86 may be reached. This subroutine is responsible for normal sampling and includes five states, zero through four. Briefly, at block 88, a sensor set-up subroutine is represented as including states zero to three of the period zero subroutine 86. As will be discussed in greater detail below, during these states sensor parameters including amplifier gain A and offset voltages V_{OS} are initialized, provided that a finger is present in the sensor. State four of the interrupt period zero subroutine 86 is the normal data acquisition state and is shown at block 90. This state is reached when, for example, a finger is present in sensor 12, the amplifier gain, offset voltages, and driver currents are within their appropriate ranges, and the software is not performing a test task. As shown in FIGURE 16, state four of the period zero subroutine 86 includes a number of instructions. At block 92 drive currents are applied to LEDs 40 and 42 and the resulting signal produced by detector 38 is sampled. The signals produced in response to light at each wavelength are then compared against the desired operating ranges to determine whether modifications of the driver currents and voltage offsets are required. This step is shown at block 94. The exact manner in which these control variables are tested and modified is discussed above and in greater detail below. Finally, as shown in block 96, state four of the period zero subroutine 86 updates the displays 20 of oximeter 10.

The operation of the period zero subroutine 86 of the interrupt routine 82 is now discussed in greater detail in conjunction with FIGURE 17. As shown at block 98, the current state of sensor set-up is determined, including the level of LED drive current, amplifier gain, and offset voltages. At block 100, these sensor set-up states are analyzed to determine which of the five period zero interrupt state subroutines, indicated by blocks 102, 104, 106, 108, and 110, are to be executed. The manner in which the various state subroutines are sequenced for execution by block 100 is shown in greater detail in FIGURE 18. Generally, states are entered in ascending order. Thus, the state zero subroutine 102 forms an initial block 112 in the sequencing performed by block 100. In this state, as noted above and discussed in greater detail below, the various parameters and variables employed by oximeter 10 are initialized. As shown, the sequential processing returns to state zero whenever the conditions required for a particular state routine are violated. Given the general processing of state routines in ascending order, with the parameters initialized at block 112, the state one subroutine 104 is reached at block 114. This routine sets the drive currents applied to LEDs 40 and 42. The state one subroutine 104 is maintained until the drive currents are set, at which time sequential processing moves to the state two subroutine 106 shown at block 116. In the event that the state one subroutine 104 determines that the LED drive currents cannot be set, processing is returned to the state zero subroutine at block 112.

Once the state two subroutine 106 of the interrupt period zero subroutine 86 is reached at block 116, the offset voltages are adjusted. The state two subroutine 106 is maintained until the offsets are properly adjusted or it is determined that they cannot be so adjusted given the current drive settings. With the offsets properly adjusted, sequential processing continues to the state three subroutine at block 118. If they cannot be properly adjusted, however, the interrupt period zero subroutine 86 is reset to state zero.

The amplifier gain levels determined during calibration are set in the state three subroutine shown at block 118. Once properly set, sequencing continues to the state four subroutine at block 120, where the normal analog signal processing is performed.

With this basic understanding of the various states of the period zero normal sample subroutine 86 and the sequential order in which those states are processed, a more detailed discussion of the various states of the period zero subroutine is now provided. FIGURE 19 is a detailed flow chart of the processing included in the state zero subroutine 102 of FIGURE 17. As shown in block 122, the first instruction in the state zero subroutine 102 calls for normal signal processing to be halted. At block 124, the gain for each channed is reset to one and the offset voltage for each channel is reset to zero at block 126. Similarly, the LED drive currents for each channel are initialized to their maximum values at block 128. With these conditions performed, the sequential processing of the various state routines of the period zero interrupt subroutine 88 cause the state one subroutine 104 to be reached at block 130. A return is provided at block 132 and the processing associated with the state one subroutine will occur at the time of the next period zero interrupt.

The state one subroutine 104 of FIGURE 17 is shown in greater detail in FIGURE 20. When the state one subroutine 104 is executed, a pulse search flag is set at block 107 to indicate that no pulse is available. At block 109, the presently established drive currents are output to LEDs 40 and 42 and the resulting signals produced from detector 38 are read. The signals input to the drivers 44 are then checked to determine whether they are in a valid operating range at block 111. More particularly, a test is performed at block 113 to determine whether the driver inputs are below some predetermined acceptable minimum value. If either input is below the acceptable minimum, the output of the corresponding driver 44 is then tested at block 115 to determine whether it is at some predetermined maximum value. With the output at a maximum, it is assumed that no greater drive current can be provided and the interrupt period zero subroutine 86 is advanced to state two at block 117 via block 119.

If, on the other hand, the input to either driver 44 is determined to be less than the predetermined minimum at block 113, but block 115 indicates that the drive current is not at its maximum value, the routine progresses via block 121 to block 123 where the drive is increased in proportion to the drive current read. In this manner, an attempt is made to adjust the drive current to a nominally desired level. More particularly, the input to driver 44 is increased by an amount equal to some nominal value minus the present input, all divided by a predetermined constant. The validity of the drive setting is then checked the next time that interrupt period zero routine 86 is performed.

If block 113 determines, for each channel, that the input to driver 44 is not below the predetermined minimum, a test is performed at block 125 to determine whether the input is above some predetermined maximum. If an input signal is above the desired range, the output of the corresponding driver 44 is checked at block 127 to determine whether it is at some predetermined minimum value. A driver output at its minimum value is interpreted at block 129 to indicate that no finger is present in the sensor and the routine is returned to state zero. If, on the other hand, block 127 indicates that the output of driver 44 is not at its mininum value, block 123 is reached via block 121, allowing the drive output to be adjusted in an attempt to bring the drive input back below the maximum predetermined value exceeded at block 125. More particularly, the drive output is adjusted at block 123 by decreasing the input to driver 44 by an amount equal to a nominal input minus the present input divided by some predetermined factor.

Finally, if the driver input falls within a predetermined desired operating range, the tests at block 113 and 125 will both be failed and the subroutine will progress to block 117 via block 119. This indicates that the drives are properly set and block 117 allows the subroutine to progress to state two at the next period zero interrupt via block 131. Although not shown in FIGURE 20, such valid drive input values must exist multiple concurrent times before the state two subroutine 106 is reached.

FIGURE 21 is a pictorial illustration of the response of the state one subroutine 104 to the various input and output levels of the drivers 44. As shown, if the input current to driver 44 is below a predetermined minimum, DI1, an invalid condition is indicated unless the output current of driver 44 is greater than some predetermined maximum DO1. If DO1 is not exceeded, the driver inputs are adjusted upward before the state two subroutine 106 is reached. Similarly, if the input signal to drivers 44 exceeds some predetermined maximum DI2, an invalid condition is indicated and the input to the driver is adjusted downward. When the input signal to drivers 44 falls between the levels DI1 and DI2, however, the signal is within a valid range and the state two subroutine 106 is directly accessed.

FIGURE 22 is a more detailed flow chart of the state two routine 106 of the period zero interrupt subroutine 86. As shown, at step 133, the drive currents previously established at the state one subroutine 104 are provided to LEDs 40 and 42 and the response of detector 38 as processed by I/O circuit 14 is read. Thus, a signal is received corresponding to each wavelength of light and the signals of both channels are processed in the following manner. At block 134, the signal at each channel is checked to determine whether it is within an absolute range extending between limits L1 and L2 as shown in FIGURE 15. As indicated at block 136, if the signal is outside this predetermined range, the state is reset to zero at step 138 and the input signal to drivers 44 is adjusted in the manner described above to produce an acceptable drive current for LEDs 40 and 42.

When the test performed at block 136 indicates that the signals from A/D converter 72 are within the range between L1 and L2, the state two subroutine 106 progresses to block 140. There, the signal for each channel is checked against a window defined between an upper limit L5 and a lower limit L6. If the signal falls within those limits, block 142 of the subroutine causes the program to be advanced to state three of the period zero interrupt subroutine 86 sat block 144 without further adjustments being made. This indicates that the offset voltages are properly set. If, on the other hand, block 142 indicates that the signal is not within the window defined between L5 and L5, block 146 adjusts the offset voltage employed by subtractor 66 in proportion to the level of the signal received. For example, the offset may be either increased or decreased by an amount equal to the signal level divided by some predetermined factor. The adjustment of the offset voltage is accomplished by establishing a new offset code to be transmitted to subtractor 66. This new code is checked at block 148 to determine whether it is within a range of valid codes. The test is performed at block 150 and if the new code is not valid, the state is reset to zero via block 138 so that the drive currents for LEDs 40 and 42 can be reinitialized. If the code is in range, a return is assessed at block 152.

The operation of the state three subroutine 108 of the period zero interrupt subroutine 86 is shown in greater detail in FIGURE 23. At block 154, the present drive current signals are provided to drivers 44 to drive LEDs 40 and 42. No samples of the input channels from I/O circuit 14 are made in this state. At block 156, a flag is set to initiate signal processing. In addition, a flag is set at step 158 to cause the generation of an appropriate independently derived calibration curve based upon information received from sensor 12. A gain code established during calibration is then set at block 160 to determine the gain of amplifier 68 and the period zero interrupt routine 86 is progressed to the state four subroutine 110 at block 162 via return 164.

The details of the state four subroutine 110 are shown in FIGURE 24. Processing begins with the output of signals to drivers 44 for the production of drive currents at LEDs 40 and 42. Signals from each channel as processed by the subtractor 66 and amplifier 68 are then sampled and stored at block 166. These samples are then checked at block 168 against the minimum and maximum range limits L3 and L4 shown in FIGURE 15. If the signal is not within those limits, block 170 causes the subroutine to progress to block 172 where an indication is produced that a finger is no longer present in sensor 12 and the state is reset to zero via step 174.

If, on the other hand, the signal is within the absolute maximum range, the samples are again checked at block 176 to determine whether they are within the desired operating range, shown in FIGURE 15 as lying between limits L5 and L6. If they are, block 178 directs the program to a return at block 180. This indicates that the samples received are acceptable and allows computational software to produce a value of R_{OS}. If block 178 indicates that the signal is not within the desired operating range, however, the program is directed to step 182 where the offset voltage is adjusted upward or downward by a factor of eight to bring it within the desired range. At block 184, the code for this new offset voltage is output.

As noted previously, before the signal samples produced in the state four subroutine 110 can be used in computations, they must be converted back to valid signals that do not include the effects of gain and offset. The instructions 186 for this reconstruction process are shown in FIGURE 25. This routine is executed during an interrupt event period not shown in FIGURE 16. Signal reconstruction is performed for each channel and begins at a first block 188 where the sign of the offset code currently in use is determined. Then, at block 190, the absolute value of the offset code is extracted and divided by four at block 192. This divided offset code is used as an index for a calibrated offset table shown in FIGURE 26. This table is generated during calibration and contains the calibrated offset voltage corresponding to each offset code.

From block 192, the signal reconstruction routine 186 progresses to block 194 where the present signal sample for the channel involved, as produced by the period zero interrupt subroutine 86, is retrieved. At block 196, the equivalent zero reference value, determined at calibration, is subtracted from the input sample to produce a signed value. To allow subsequent arithmetic operations to be performed with the retention of a greater number of bits, the 12-bit input signal is scaled to a 16-bit number at block 198.

At block 200, the gain previously applied to the signal by amplifier 68 is removed. The effect of the offset voltages is them removed from the signal in the following manner. The calibrated offset value equivalent to the offset code is extracted from the table shown in FIGURE 26 at block 204. This calibrated offset value is converted to its signed equivalent at block 206 and is subtracted from the previously processed signal at block 208. This value is then stored at block 210 for subsequent processing accomplished via a return at block 212.

In addition to the analog sample processing task discussed above, the software may include a time task, display drive task, keyboard operation task and test routine task. The analog sample processing task has the highest priority of these various tasks.

As noted, the instructions for the software that controls the signal construction-reconstruction process discussed above are stored in EPROM 78 of micropùter 16. Similarly, values for R_H, R_L, IR_H, IR_L, and signal period are determined pursuant to peak-detection software contained in EPROM 78. These values are stored in random-access memory (RAM) 214 for operation upon by CPU 80 in accordance with further computational instructions stored in EPROM 78. Interfaces 216 act as input and output buffers for microcomputer 16.

The computational software in EPROM 78 initially causes CPU 80 to determine the present value for R_{OS} by substituting the measured values for R_H, R_L, IR_H, and IR_L into equation (22):

$$R_{OS} = \frac{\ln (R_L/R_H)}{\ln (IR_L/IR_H)} \qquad (26)$$

Then, the computational software instructs CPU 80 to determine the oxygen saturation from R_{OS} by use of a calibration curve, such as the one depicted in FIGURE 7. The calibration curve is a plot of the relationship between independently determined oxygen saturations corresponding to values of R_{OS}produced by oximeter 10 in accordance with the technique described above.

With sufficiently large space in EPROM 78, enough points along the calibration curve can be stored in a look-up table to allow CPU 80 to extract an accuate indication of oxygen saturation from the value of R_{OS} input to EPROM 78. The storage of a sufficient number of calibration curve data points may, however, necessitate the use of an undesirably large-capacity EPROM 78. For that reason, a second method of storing the calibration curve information is preferred.

Pursuant to that method, once independently derived data associating R_{OS} with the oxygen saturation is obtained, a mathematical expression between the two can be derived from a plot of the curve. The basic formula and the coefficients of the formula's variables are then stored in EPROM 78. When a value for R_{OS} is measured, CPU 80 extracts the coefficients from EPROM 78 and computes a value for the oxygen saturation. This technique allows information completely identifying the entire calibration curve, or a family of such curves, to be stored within a relatively small amount of EPROM 78 space.

The computational software in EPROM 78 also instructs CPU 80 to determine the pulse rate from the signal period. Displays 20 then provide visible and audible outputs of the oxygen saturation and pulse rate in a manner conveniently used by the operator of oximeter 10.

While the references have been described with reference to a preferred embodiment, it is to be clearly understood by those skilled in the art that the invention is not limited thereto, and that the scope of the invention is to be interpreted only in conjunction with the following claims.

Claims

1. An apparatus for receiving and processing signals, produced by a sensor (12), that contain information about the oxygen saturation of arterial blood flowing in tissue, said apparatus characterized by:
offset subtraction means (66) for subtracting from said sensor means signal a controlled portion of said signal, said offset subtraction means (66) having an output that is substantially equal to the portion of said sensor signal remaining after said controlled portion is substracted therefrom;
control means (16) for receiving said output of said offset subtraction means (66) and producing a subtraction control signal that is dependent on said output, said subtraction means (66) receiving said

subtraction control signal and adjusting said control portion of said sensor signal in response thereto; and analyzing means (16) for receiving said output of said subtraction means (66) and producing an indication of the oxygen saturation of said arterial blood flowing in said tissue.

2. The apparatus of Claim 1, wherein said subtraction control signal produced by said control means (16) causes said offset subtraction means (66) to continue subtracting the same said controlled portion from said sensor signal when the value of said subtraction means output is within a first predetermined range.

3. The apparatus of Claim 2, wherein said subtraction control signal produced by said control means (16) causes said offset subtraction means (66) to adjust the magnitude of said controlled portion of said sensor signal when the value of said subtractions means output falls within a second predetermined range, said magnitude of said controlled portion being adjusted in proportion to the magnitude of said subtraction means output.

4. The apparatus of Claim 3, wherein said subtraction control signal produced by said control means (16) indicates that said offset subtraction means (66) is unable to adjust said controlled portion of said sensor signal when the value of said subtraction means output exceeds a third predetermined level.

5. The apparatus of Claim 4, wherein said controlled portion of said signal subtracted by said offset subtraction means (66) is initialized at a predetermined value.

6. The apparatus of Claim 1, wherein said analyzing means (16) adds said controlled portion of said signal back to said offset subtraction means output before producing said indication of oxygen saturation.

7. The apparatus of Claim 1, further comprising an amplifier (68) for amplifying said subtraction means output by a controlled gain before said subtraction means output is received by said analyzing means (16), said amplifier (68) having an output that is substantially equal to the product of said subtraction means output and said gain, said control means (16) producing an amplifier control signal that is received by said amplifier (68), which adjusts the controlled gain in response thereto.

8. The apparatus of Claim 7, wherein said control means (16) produces a sensor control signal to which said sensor (12) responds.

9. The apparatus of Claim 1, wherein said control means (16) produces a sensor control signal to which said sensor responds.

10. The apparatus of Claim 9, wherein said control means (16) establishes said sensor control signal at a level sufficient to cause said signal produced by said sensor (12) to fall within a predetermined sensor signal range.

11. The apparatus of Claim 10, wherein said sensor control signal produced by said control means (16) causes said sensor (12) to adjust the level of said signal when said subtraction means output is within a predetermined range.

12. A method of processing information signals that contain information about the oxygen saturation of arterial blood flowing in tissue, characterized by:
subtracting from said information signal a controlled portion of said information signal, the magnitude of said controlled portion being determined by a subtraction control signal;
producing a subtraction output that is substantially equal to the portion of said information signal remaining after said controlled portion is subtracted therefrom; and
producing said subtraction control signal, the magnitude of said subtraction control signal produced being a function of said subtraction output and indicating the desired adjustment in said controlled portion subtracted from said information signal.

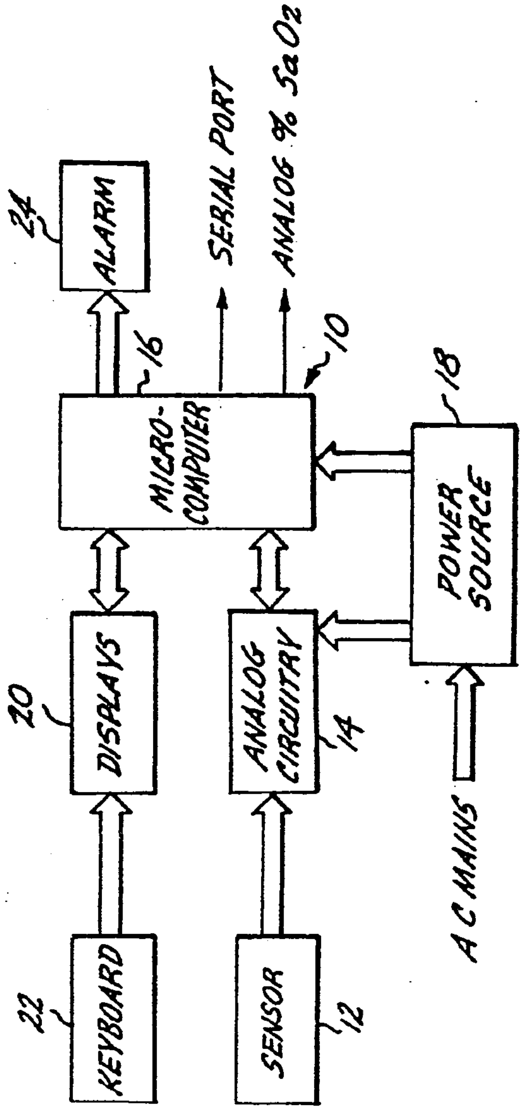
22
KEYBOARD
12
SENSOR
20
DISPLAYS
ANALOG CIRCUITRY
14
MICRO-COMPUTER
16
10
24
ALARM
SERIAL PORT
ANALOG % SaO2
18
POWER SOURCE
A C MAINS

Fig. 1.

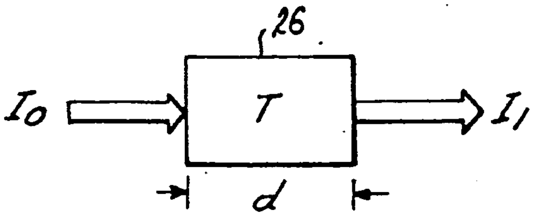

Fig.2.

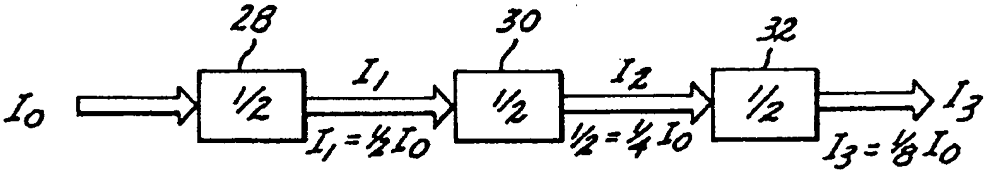

Fig.3.

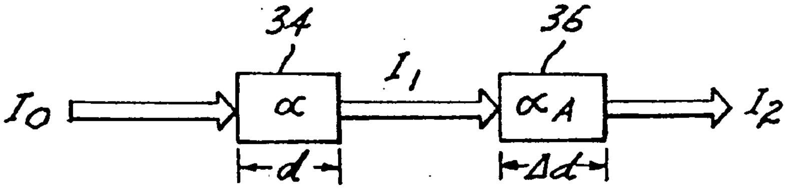

Fig.6.

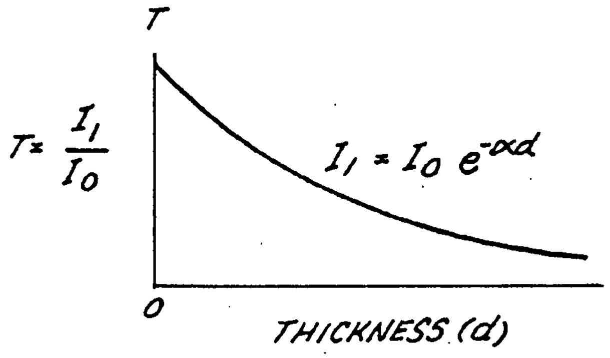

Fig. 4.

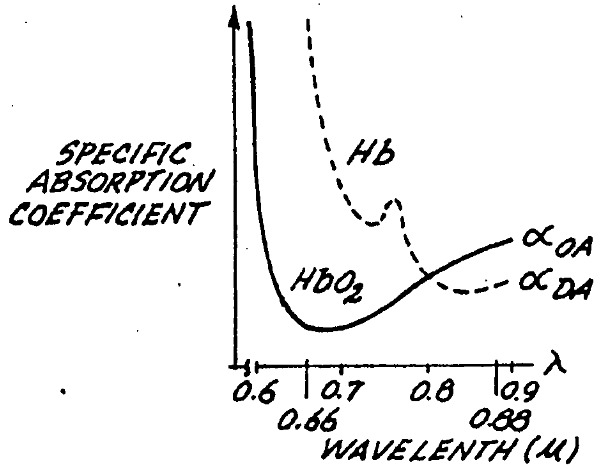

Fig. 5.

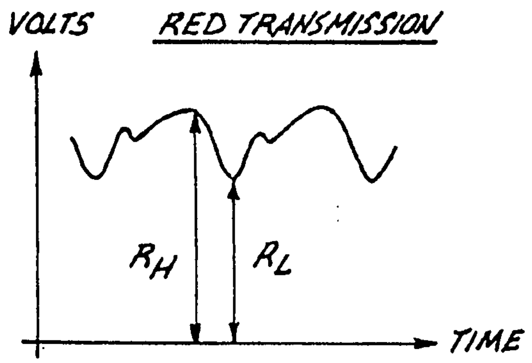

Fig. 9.

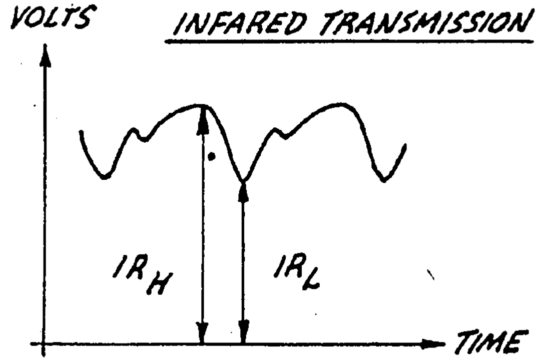

Fig. 10.

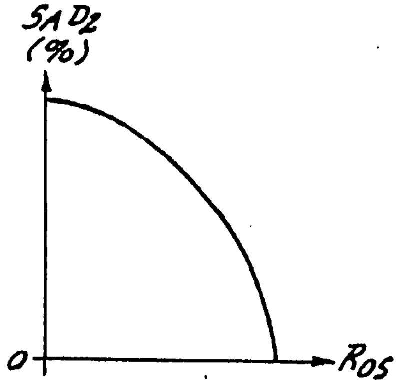

Fig. 7.

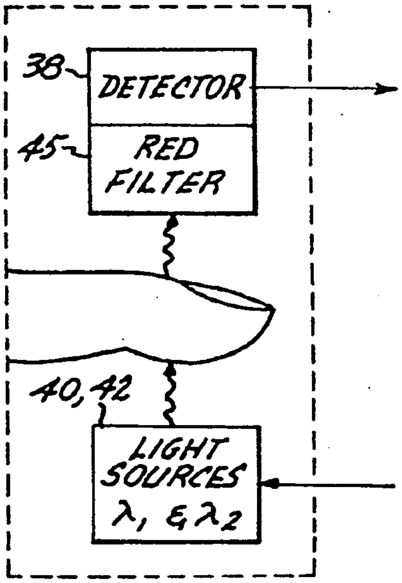
38
DETECTOR
45
RED FILTER
40,42
LIGHT SOURCES λ_1 & λ_2

Fig. 8.

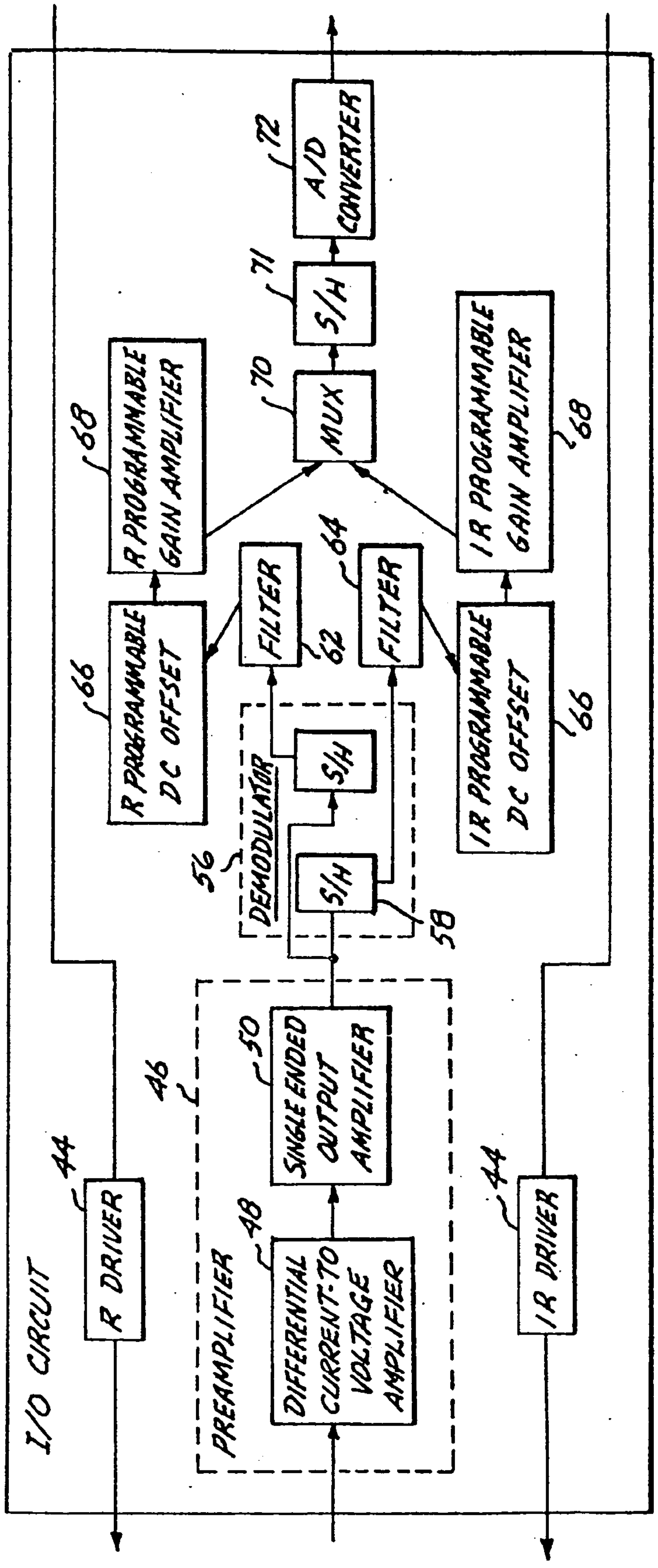
I/O CIRCUIT
R DRIVER
44
PREAMPLIFIER
46
DIFFERENTIAL CURRENT-TO VOLTAGE AMPLIFIER
48
SINGLE ENDED OUTPUT AMPLIFIER
50
I R DRIVER
44
DEMODULATOR
56
S/H
S/H
58
FILTER
62
FILTER
64
R PROGRAMMABLE DC OFFSET
66
R PROGRAMMABLE GAIN AMPLIFIER
68
I R PROGRAMMABLE DC OFFSET
66
I R PROGRAMMABLE GAIN AMPLIFIER
68
MUX
70
S/H
71
A/D CONVERTER
72

Fig.11.

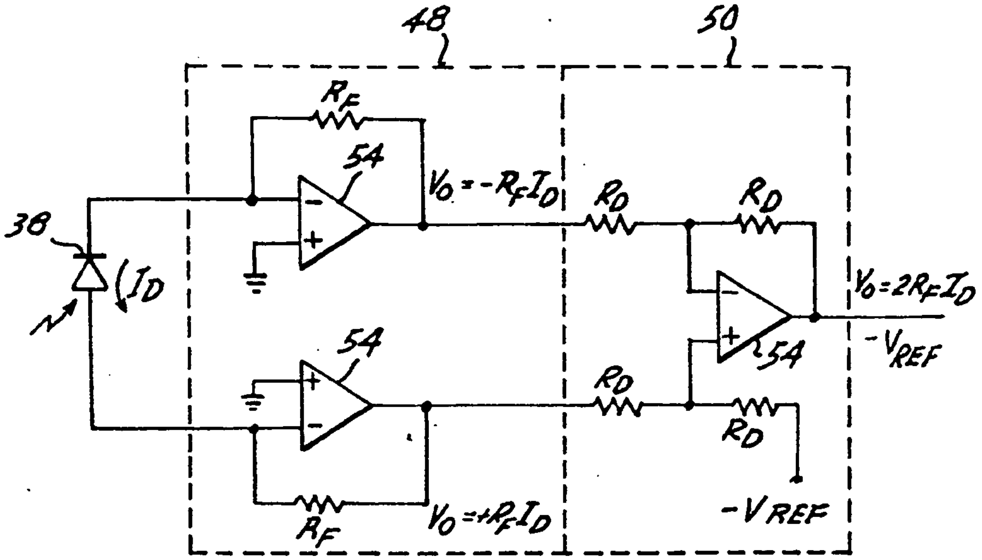
48
50
R_F
54
$V_0 = -R_F I_D$
R_D
R_D
38
I_D
$V_0 = 2R_F I_D$
54
$-V_{REF}$
54
R_D
R_D
$V_0 = +R_F I_D$
R_F
$-V_{REF}$

Fig.13.

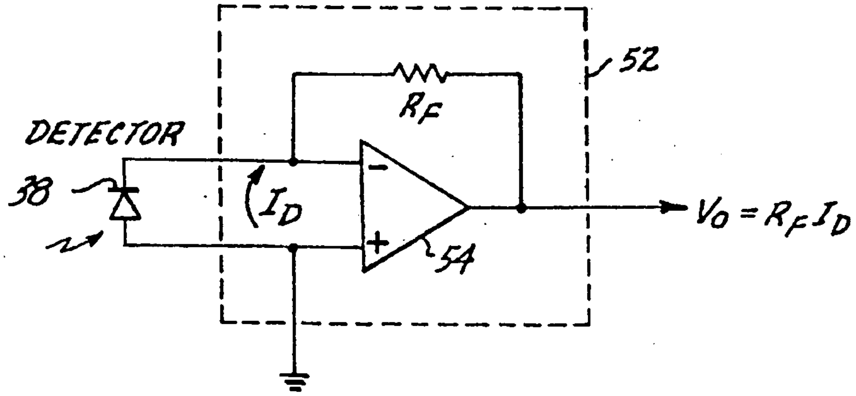
52
R_F
DETECTOR
38
I_D
54
$V_0 = R_F I_D$

Fig.12.

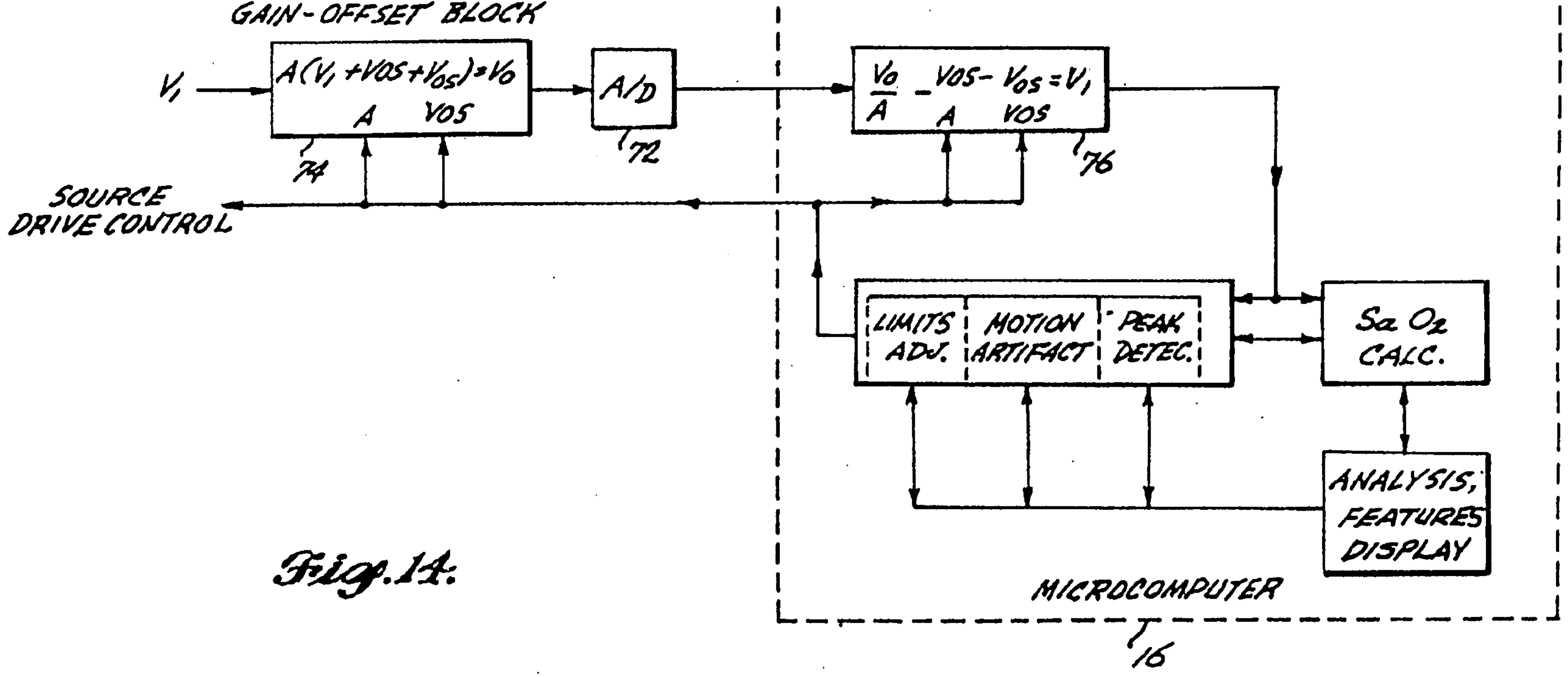
GAIN-OFFSET BLOCK
V_1
$A(V_1+VOS+V_{OS})=V_0$
A
VOS
74
A/D
72
SOURCE DRIVE CONTROL
$\frac{V_0}{A}-VOS-V_{OS}=V_1$
A
VOS
76
LIMITS ADJ.
MOTION ARTIFACT
PEAK DETEC.
SaO_2 CALC.
ANALYSIS; FEATURES DISPLAY
MICROCOMPUTER
16

Fig.14.

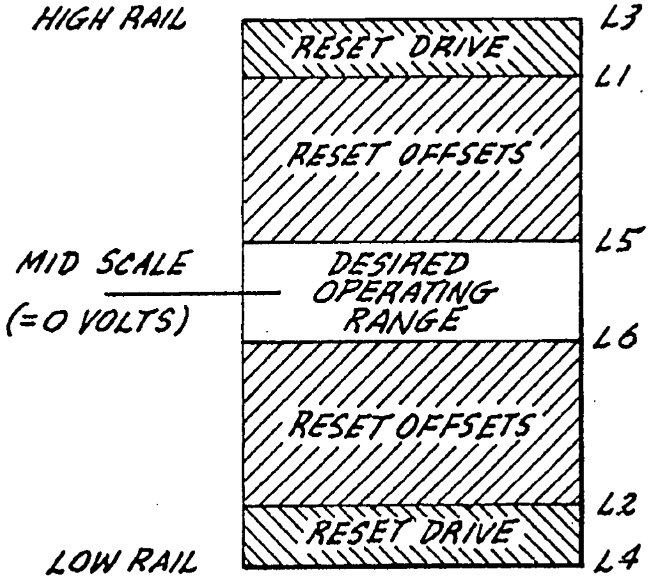
HIGH RAIL
MID SCALE
(=0 VOLTS)
LOW RAIL
RESET DRIVE
RESET OFFSETS
DESIRED OPERATING RANGE
RESET OFFSETS
RESET DRIVE
L3
L1
L5
L6
L2
L4

Fig.15.

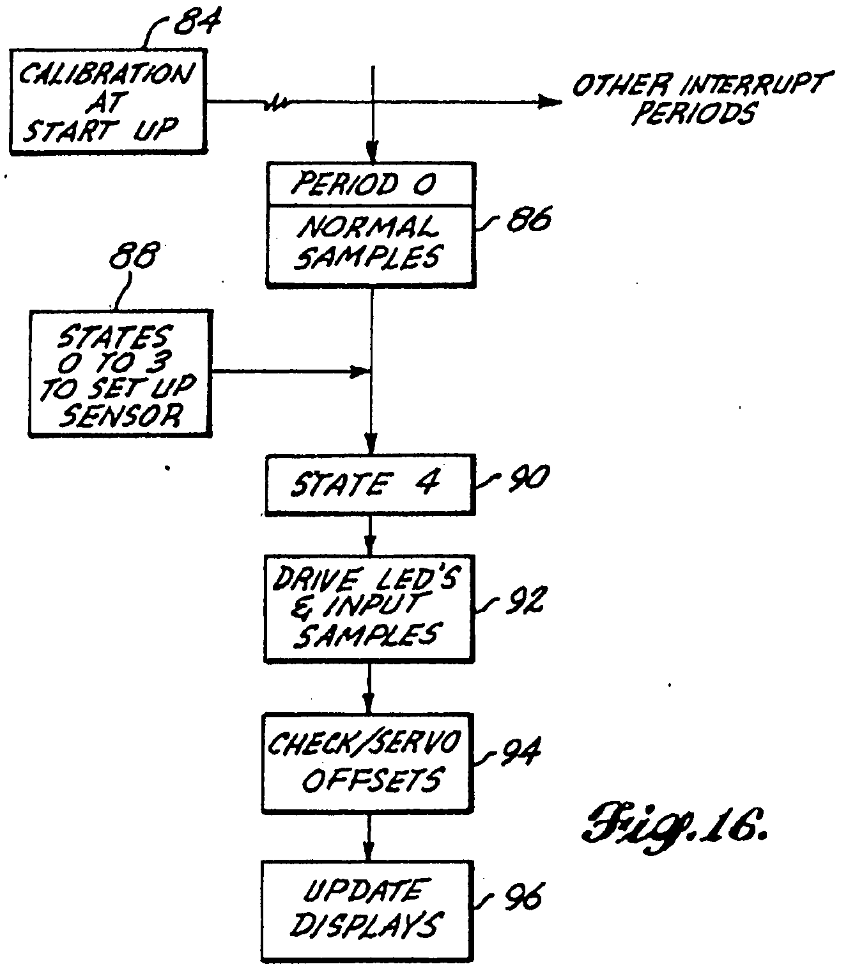
84
CALIBRATION AT START UP
OTHER INTERRUPT PERIODS
PERIOD 0
NORMAL SAMPLES
86
88
STATES 0 TO 3 TO SET UP SENSOR
STATE 4
90
DRIVE LED'S & INPUT SAMPLES
92
CHECK/SERVO OFFSETS
94
UPDATE DISPLAYS
96

Fig.16.

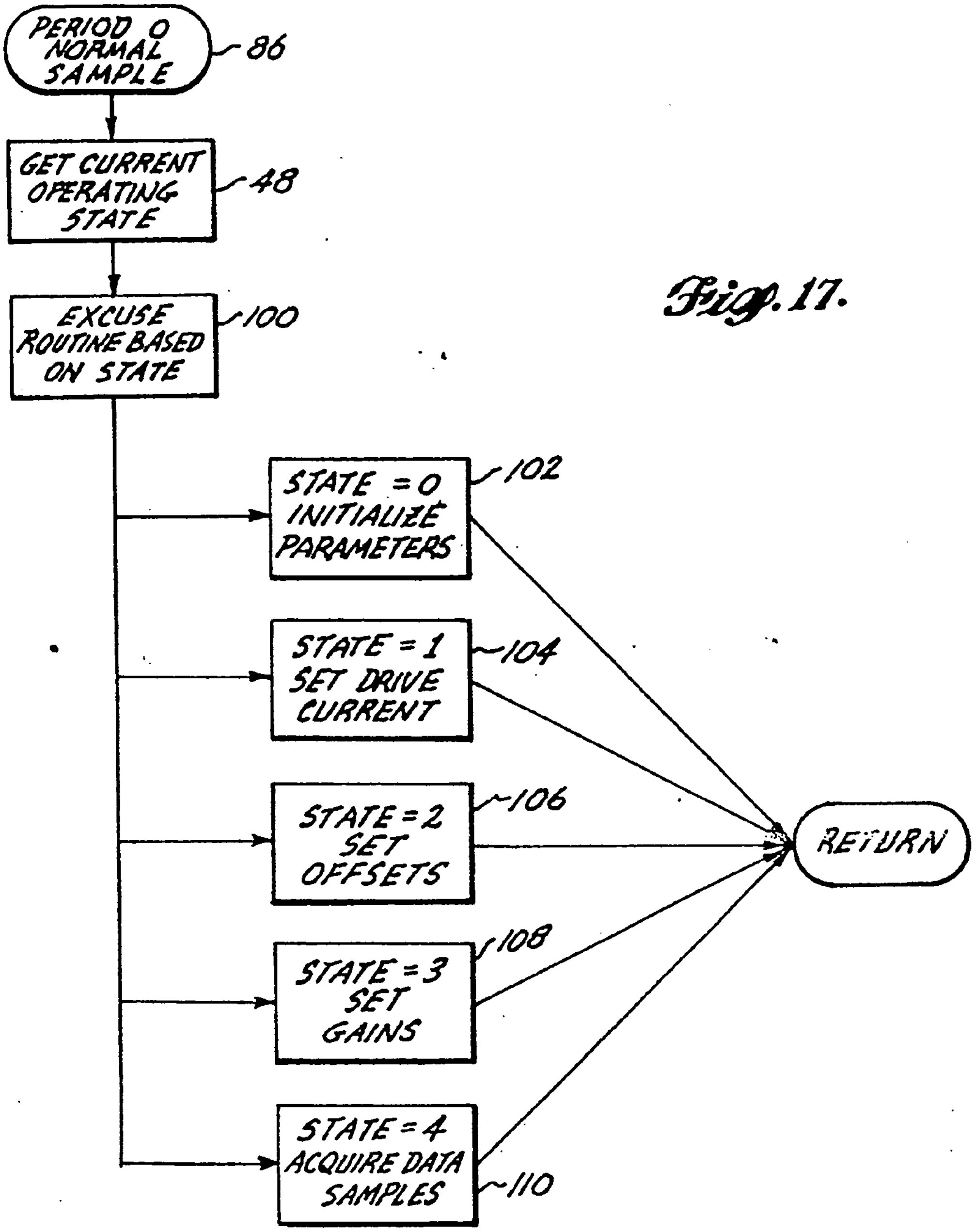
PERIOD 0 NORMAL SAMPLE
86
GET CURRENT OPERATING STATE
48
EXCUSE ROUTINE BASED ON STATE
100
STATE = 0 INITIALIZE PARAMETERS
102
STATE = 1 SET DRIVE CURRENT
104
STATE = 2 SET OFFSETS
106
STATE = 3 SET GAINS
108
STATE = 4 ACQUIRE DATA SAMPLES
110
RETURN

Fig. 17.

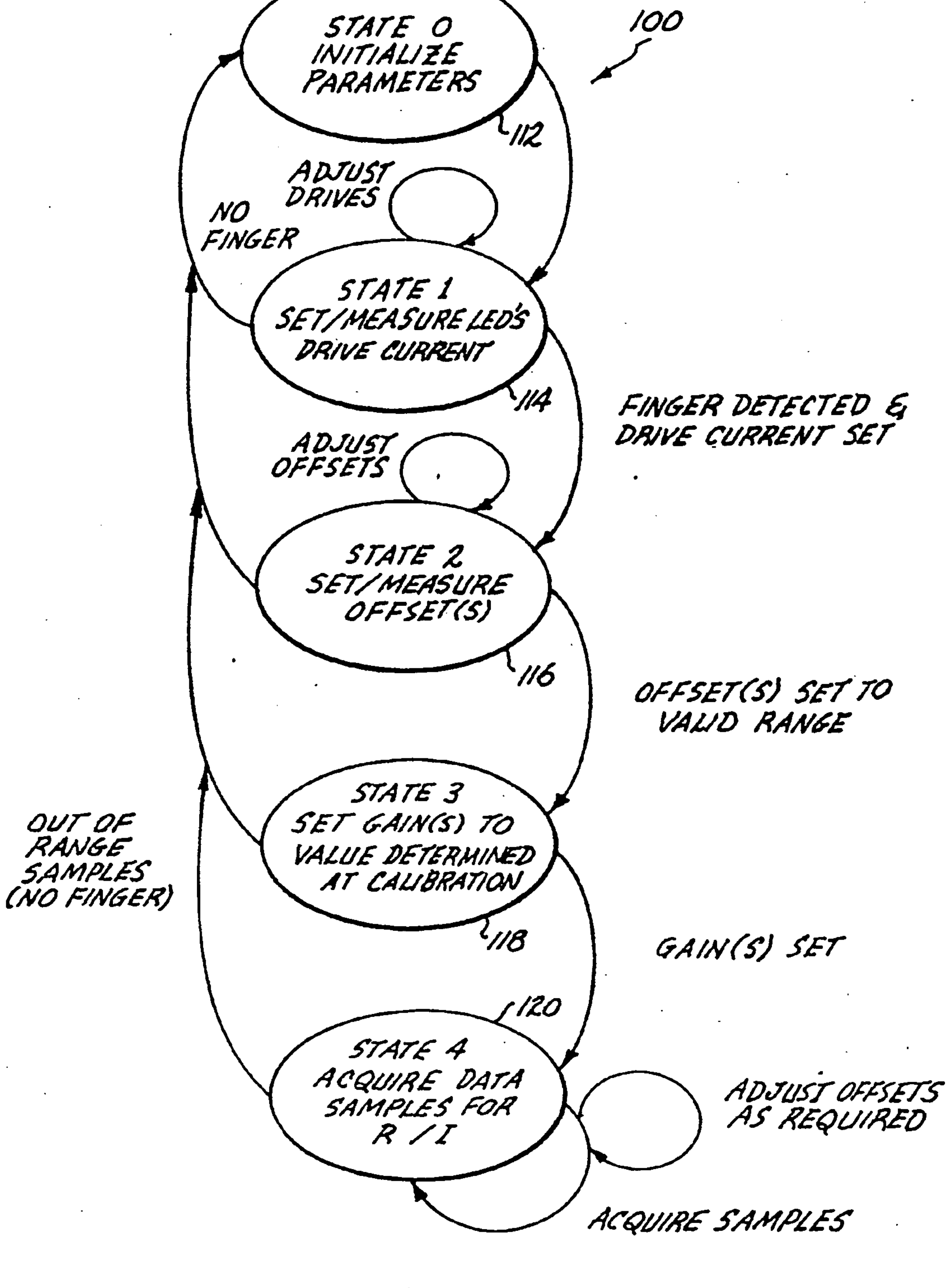
100
STATE 0
INITIALIZE
PARAMETERS
112
ADJUST
DRIVES
NO
FINGER
STATE 1
SET/MEASURE LED'S
DRIVE CURRENT
114
FINGER DETECTED &
DRIVE CURRENT SET
ADJUST
OFFSETS
STATE 2
SET/MEASURE
OFFSET(S)
116
OFFSET(S) SET TO
VALID RANGE
STATE 3
SET GAIN(S) TO
VALUE DETERMINED
AT CALIBRATION
118
OUT OF
RANGE
SAMPLES
(NO FINGER)
GAIN(S) SET
120
STATE 4
ACQUIRE DATA
SAMPLES FOR
R / I
ADJUST OFFSETS
AS REQUIRED
ACQUIRE SAMPLES

Fig. 18.

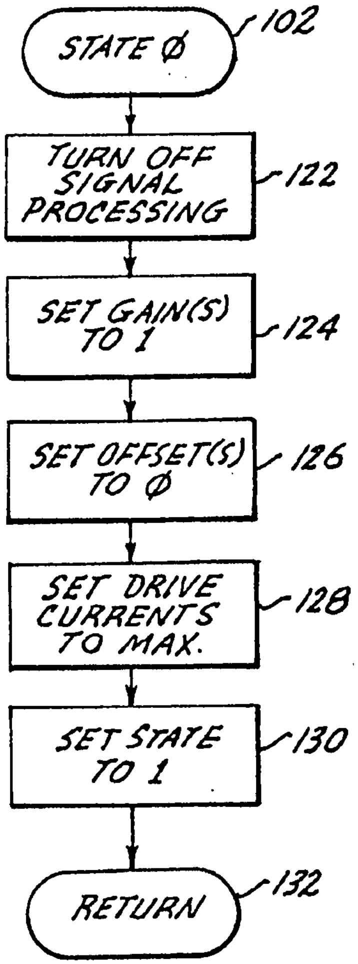
STATE Ø
102
TURN OFF SIGNAL PROCESSING
122
SET GAIN(S) TO 1
124
SET OFFSET(S) TO Ø
126
SET DRIVE CURRENTS TO MAX.
128
SET STATE TO 1
130
RETURN
132

Fig. 19.

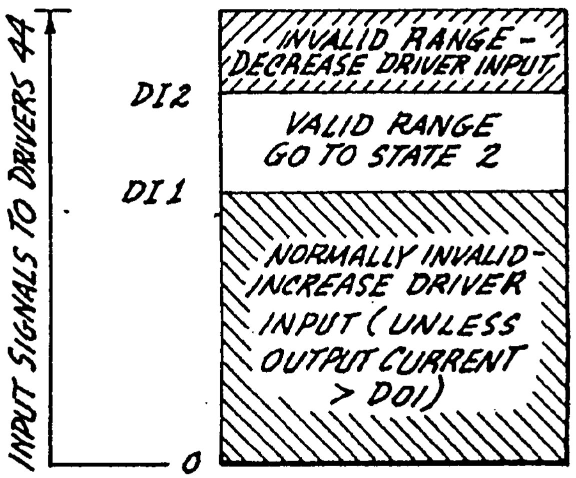
INPUT SIGNALS TO DRIVERS 44
DI2
DI1
0
INVALID RANGE - DECREASE DRIVER INPUT
VALID RANGE GO TO STATE 2
NORMALLY INVALID - INCREASE DRIVER INPUT (UNLESS OUTPUT CURRENT > DOI)

Fig. 21.

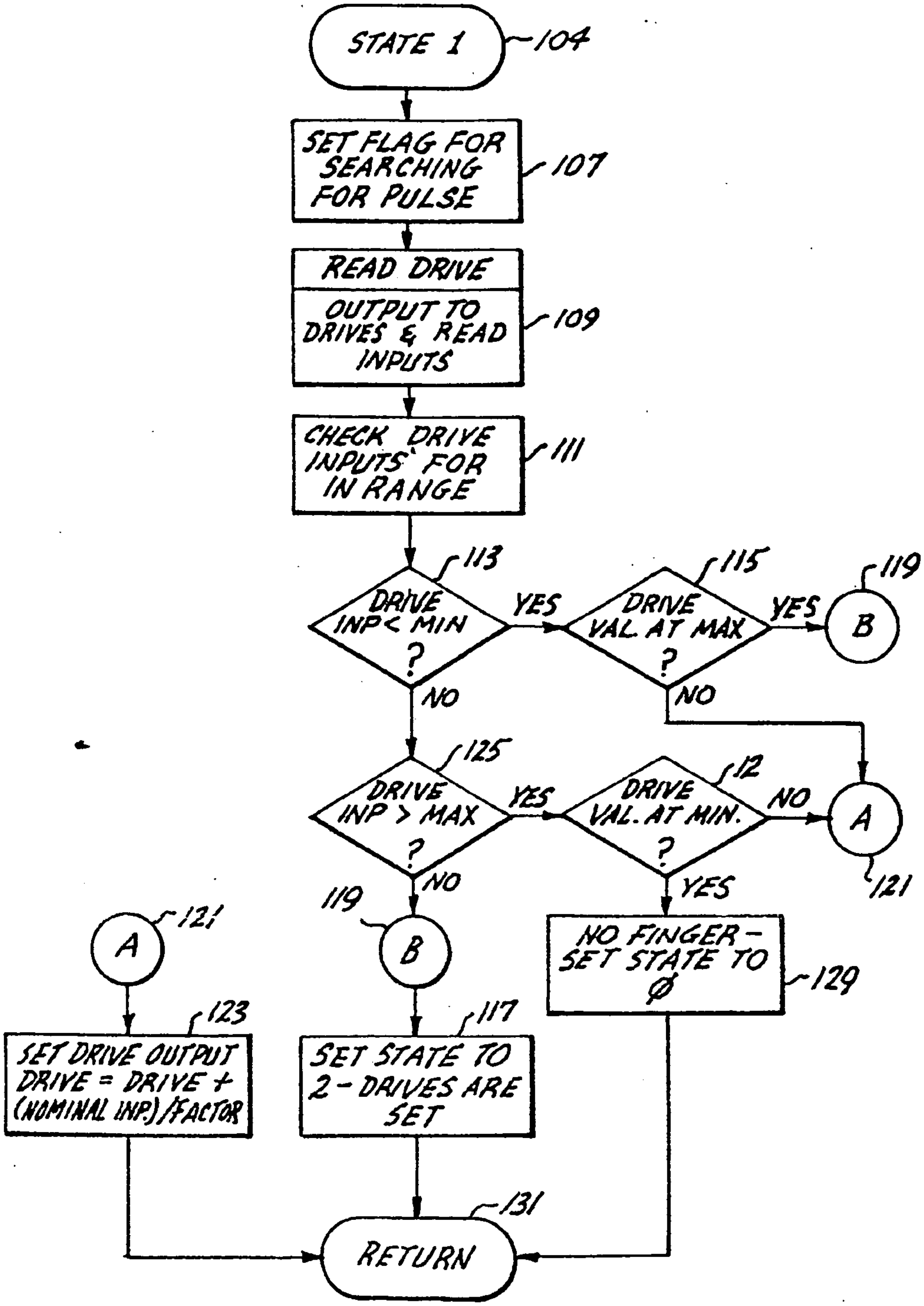
STATE 1
104
SET FLAG FOR SEARCHING FOR PULSE
107
READ DRIVE
OUTPUT TO DRIVES & READ INPUTS
109
CHECK DRIVE INPUTS FOR IN RANGE
111
113
DRIVE INP< MIN ?
YES
115
DRIVE VAL. AT MAX ?
YES
119
B
NO
NO
125
DRIVE INP > MAX ?
YES
12
DRIVE VAL. AT MIN. ?
NO
A
121
NO
YES
121
A
119
B
NO FINGER - SET STATE TO 0
129
123
SET DRIVE OUTPUT DRIVE = DRIVE + (NOMINAL INP.)/FACTOR
117
SET STATE TO 2 - DRIVES ARE SET
131
RETURN

Fig. 20.

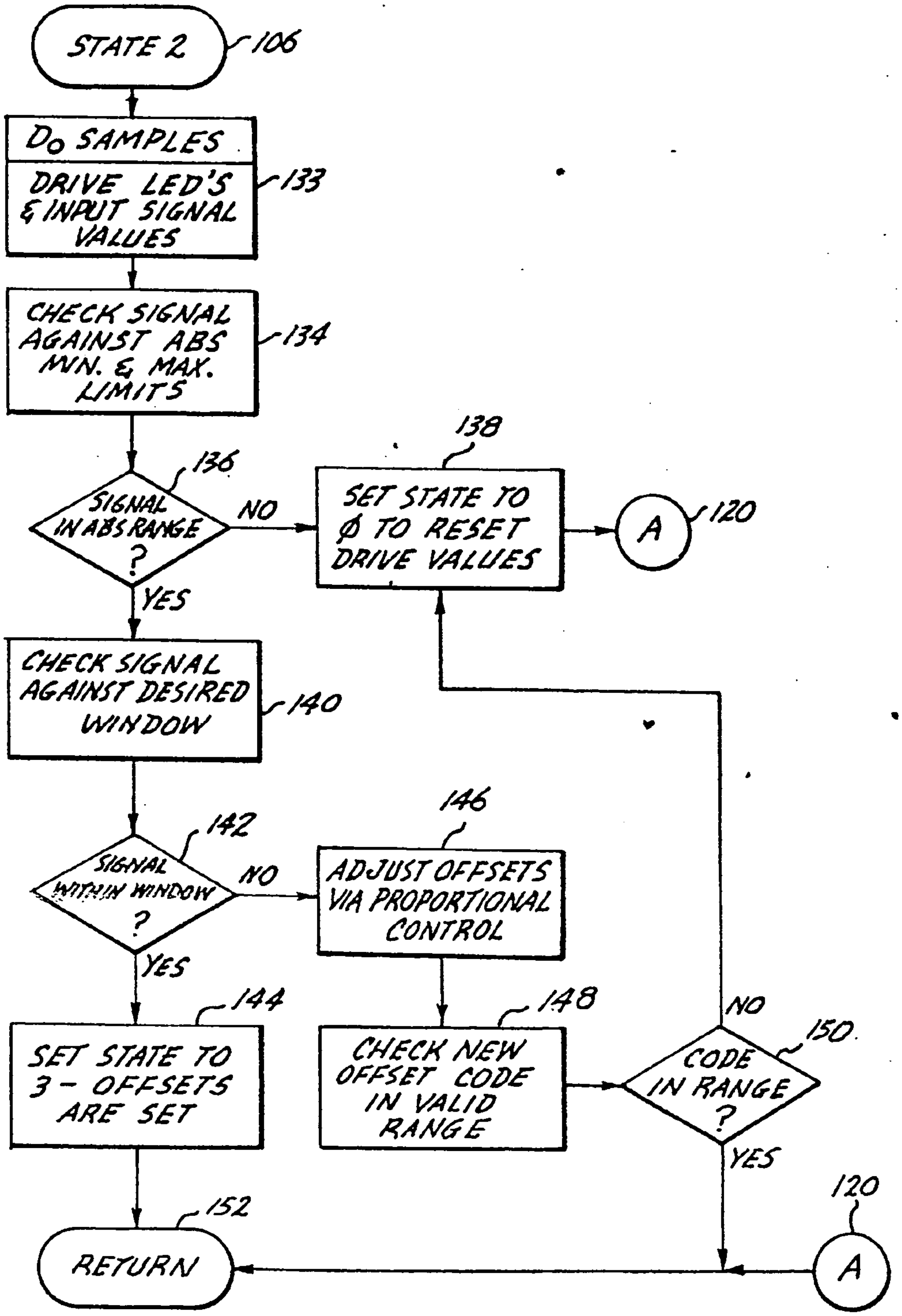
STATE 2
106
D0 SAMPLES
DRIVE LED'S & INPUT SIGNAL VALUES
133
CHECK SIGNAL AGAINST ABS MIN. & MAX. LIMITS
134
SIGNAL IN ABS RANGE ?
136
NO
YES
SET STATE TO 0 TO RESET DRIVE VALUES
138
A
120
CHECK SIGNAL AGAINST DESIRED WINDOW
140
SIGNAL WITHIN WINDOW ?
142
NO
YES
ADJUST OFFSETS VIA PROPORTIONAL CONTROL
146
SET STATE TO 3 - OFFSETS ARE SET
144
CHECK NEW OFFSET CODE IN VALID RANGE
148
CODE IN RANGE ?
150
NO
YES
RETURN
152
A
120

Fig. 22.

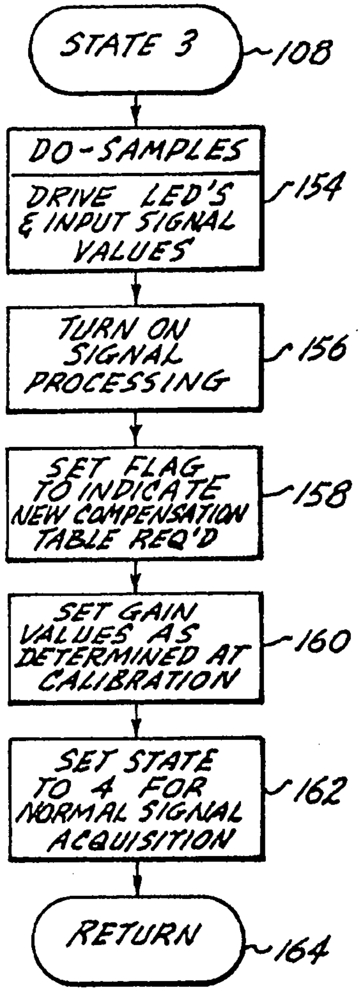
STATE 3
108
DO-SAMPLES
DRIVE LED'S & INPUT SIGNAL VALUES
154
TURN ON SIGNAL PROCESSING
156
SET FLAG TO INDICATE NEW COMPENSATION TABLE REQ'D
158
SET GAIN VALUES AS DETERMINED AT CALIBRATION
160
SET STATE TO 4 FOR NORMAL SIGNAL ACQUISITION
162
RETURN
164

Fig. 23.

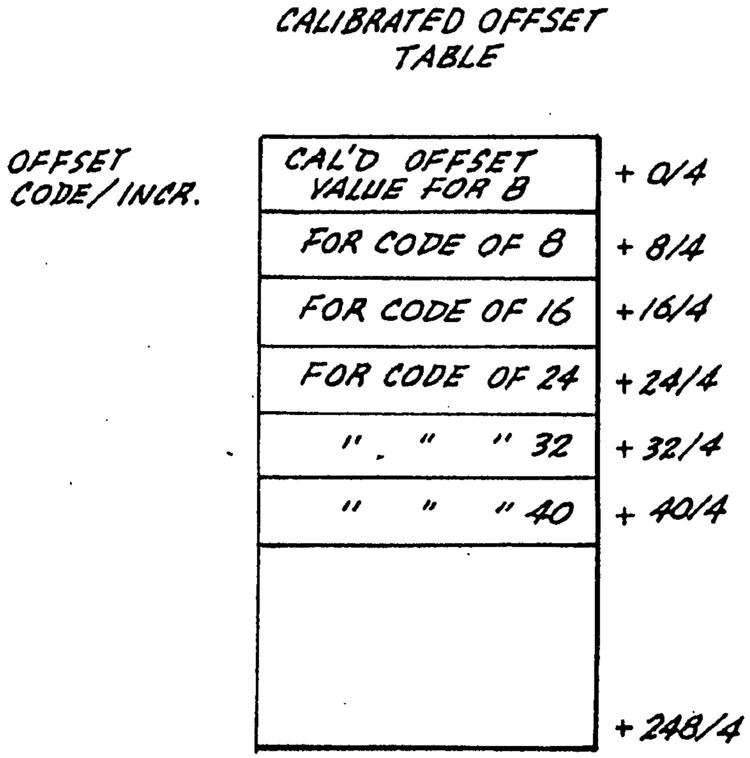
CALIBRATED OFFSET TABLE
OFFSET CODE/INCR.
CAL'D OFFSET VALUE FOR 8
FOR CODE OF 8
FOR CODE OF 16
FOR CODE OF 24
" " " 32
" " " 40
+ 0/4
+ 8/4
+ 16/4
+ 24/4
+ 32/4
+ 40/4
+ 248/4

Fig. 26.

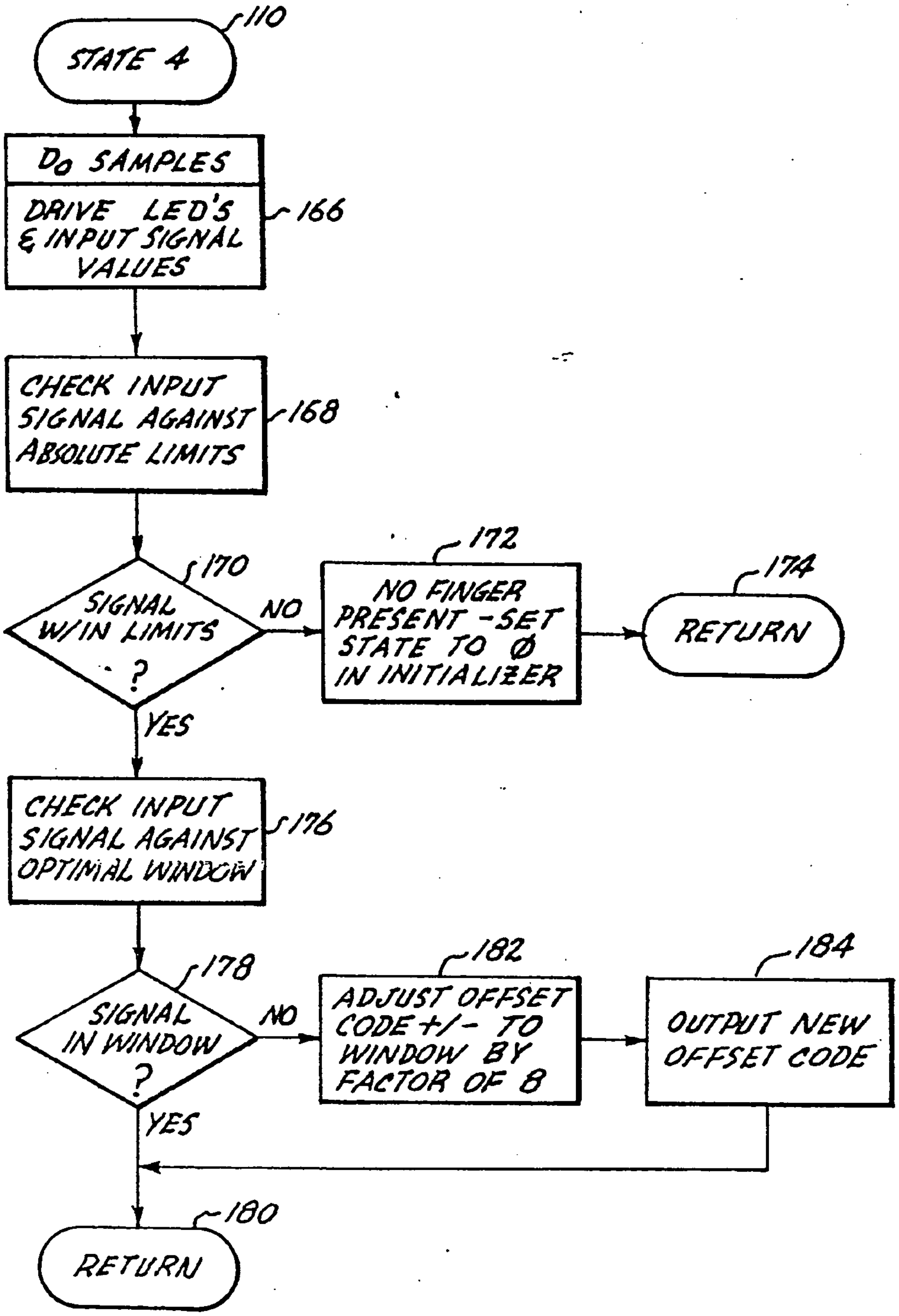
STATE 4
110
D0 SAMPLES
DRIVE LED'S & INPUT SIGNAL VALUES
166
CHECK INPUT SIGNAL AGAINST ABSOLUTE LIMITS
168
SIGNAL W/IN LIMITS ?
170
NO
YES
NO FINGER PRESENT - SET STATE TO 0 IN INITIALIZER
172
RETURN
174
CHECK INPUT SIGNAL AGAINST OPTIMAL WINDOW
176
SIGNAL IN WINDOW ?
178
NO
YES
ADJUST OFFSET CODE +/- TO WINDOW BY FACTOR OF 8
182
OUTPUT NEW OFFSET CODE
184
RETURN
180

Fig. 24.

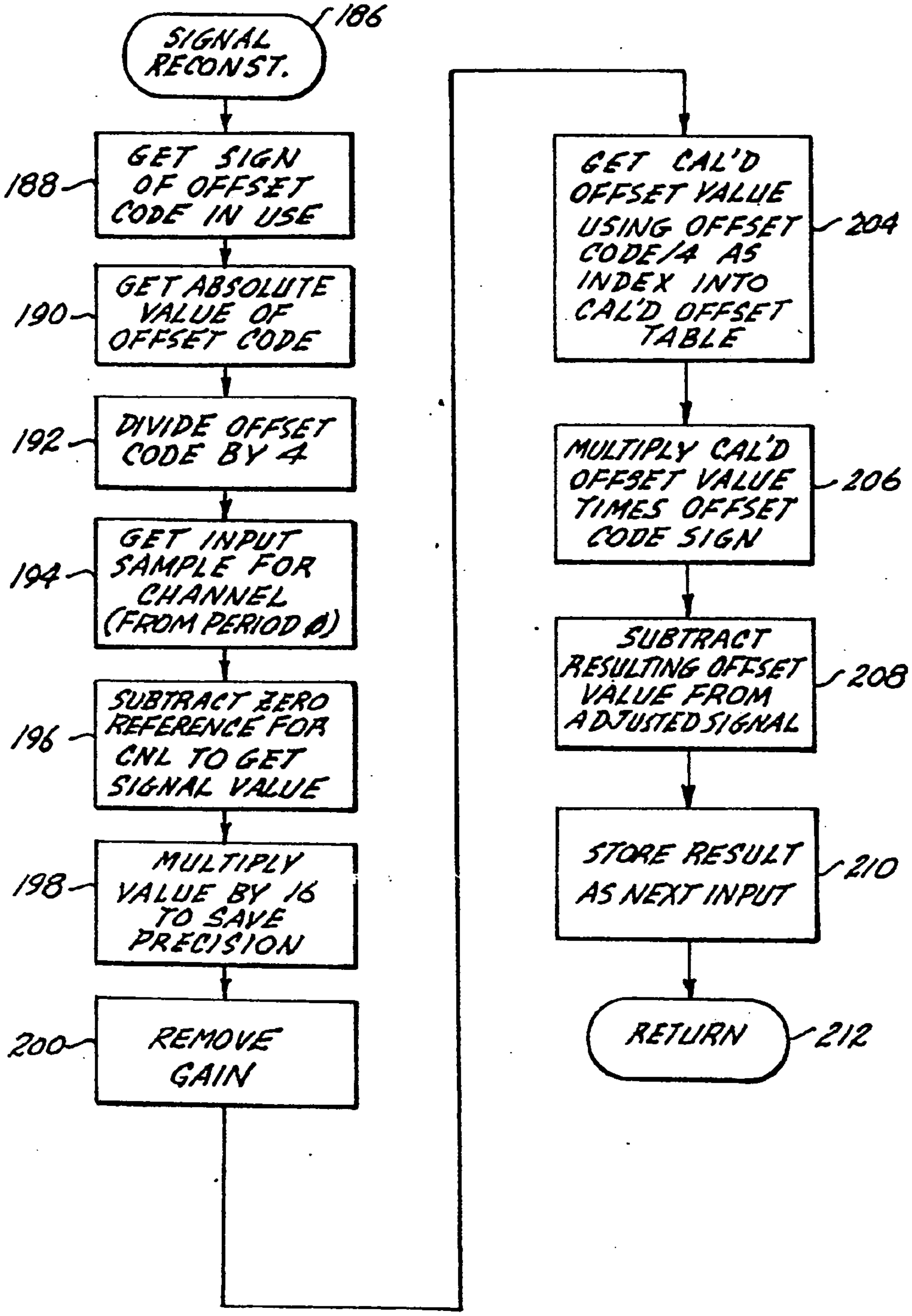
SIGNAL RECONST.
186
GET SIGN OF OFFSET CODE IN USE
188
GET ABSOLUTE VALUE OF OFFSET CODE
190
DIVIDE OFFSET CODE BY 4
192
GET INPUT SAMPLE FOR CHANNEL (FROM PERIOD 0)
194
SUBTRACT ZERO REFERENCE FOR CNL TO GET SIGNAL VALUE
196
MULTIPLY VALUE BY 16 TO SAVE PRECISION
198
REMOVE GAIN
200
GET CAL'D OFFSET VALUE USING OFFSET CODE/4 AS INDEX INTO CAL'D OFFSET TABLE
204
MULTIPLY CAL'D OFFSET VALUE TIMES OFFSET CODE SIGN
206
SUBTRACT RESULTING OFFSET VALUE FROM ADJUSTED SIGNAL
208
STORE RESULT AS NEXT INPUT
210
RETURN
212

Fig. 25.

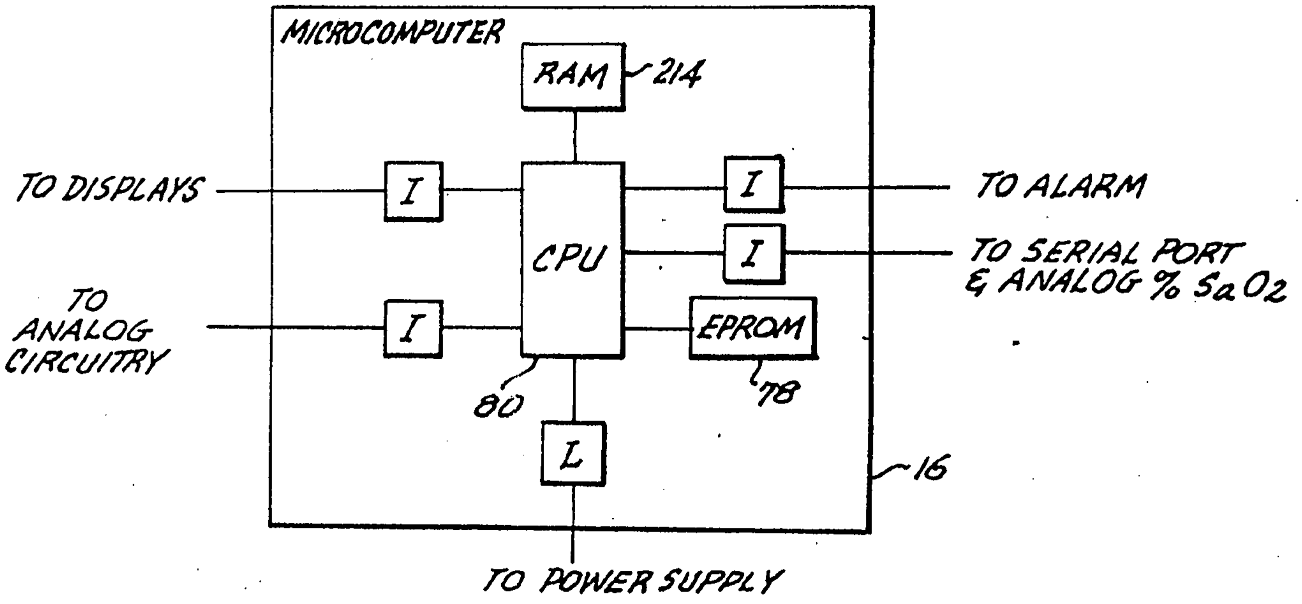
MICROCOMPUTER
RAM
214
TO DISPLAYS
I
CPU
I
TO ALARM
I
TO SERIAL PORT
& ANALOG % Sa O2
TO
ANALOG
CIRCUITRY
I
EPROM
80
78
L
16
TO POWER SUPPLY

Fig. 27.

European Patent
Office

EUROPEAN SEARCH REPORT

Application Number

EP 87 30 7258

DOCUMENTS CONSIDERED TO BE RELEVANT

Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int. Cl.4)
A	GB-A-2 151 020 (BOC) * abstract; page 1, line 116 - page 2, line 45; figure 1 * ---	1,8-10, 12	A 61 B 5/00 G 01 N 21/31
A	EP-A-0 104 772 (NELLCOR INCORP.) * page 5, line 1 - page 7, line 9; page 10, line 22 - page 11, line 16; figures 4, 6 * ---	1,8-12	
A	US-A-4 444 498 (S.O. HEINEMANN) * column 4, line 12 - column 7, line 34; figures 2A * -----	1,8-12	

TECHNICAL FIELDS SEARCHED (Int. Cl.4)

A 61 B 5/00
G 01 N 21/00
G 06 F 15/00

The present search report has been drawn up for all claims

Place of search	Date of completion of the search	Examiner
BERLIN	23-11-1987	WEIHS J.A.

CATEGORY OF CITED DOCUMENTS

X : particularly relevant if taken alone
Y : particularly relevant if combined with another document of the same category
A : technological background
O : non-written disclosure
P : intermediate document

T : theory or principle underlying the invention
E : earlier patent document, but published on, or after the filing date
D : document cited in the application
L : document cited for other reasons

& : member of the same patent family, corresponding document